

**THESIS TITLE:**

MUSCULOTENDINOUS STIFFNESS AND MUSCLE FUNCTION

**CANDIDATE:**

LAWRENCE WAYNE VILJOEN

**STUDENT NUMBER:**

VLJLAW002

**SUPERVISOR:**

ASSOCIATE PROFESSOR M.I. LAMBERT

**Thesis Presented for the Degree of**

**DOCTOR OF PHILOSOPHY**

in the Department of Human Biology  
and the Faculty of Health Sciences

**UNIVERSITY OF CAPE TOWN**

**SOUTH AFRICA**

**July 2004**

UCT/MRC Research Unit for Exercise Science and Sports Medicine

Sport Science Institute of South Africa

Boundary Road, Newlands, 7700

South Africa

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## **DEDICATIONS**

**To the Lord, God:** For providing me with this incredible opportunity to study and better myself, and for surrounding me with such amazing and truly gifted people. For watching over me, and making me grow as a person and learn so much about life.

**My Greek family: Chanté, Theo, Leonidas and Irakles:** Lots of love to you all! I dedicate this thesis to you all, being so very far away. Without family one does not have much in this world...

**My parents, Herc and Lucille Viljoen:** A special dedication to my parents, who provided me with the opportunity I would not have contemplated otherwise. When I needed a place to stay and a roof over my head to continue my studies, you were there in an instant. Thank you for being there and supporting me throughout my life, but especially during the last five years. This one is for you...

**For my beautiful fiancée Catherine:** A final dedication goes to the love of my life, for being with me since the very beginning, even when I did not know it. For loving and supporting me throughout, and still wanting to marry me after everything! You are an inspiration to me, and all who know you...

# **ACKNOWLEDGEMENTS**

**University of Cape Town:** who contributed funding towards the compilation of this thesis

- Benfara Award for 2000
- University Scholarships Committee Award for 2000
- Benfara Award for 2001
- Human Biology Departmental Scholarship for 2001
- Human Biology Departmental Scholarship for 2002
- Benfara Award for 2002
- University Scholarships Committee Award for 2002
- KW Johnston Bequest Scholarship for 2003
- Baxter for 2003 Scholarship

**Paul Beugelink and Tiny Weyers of Zest Manufacturing PTY (Ltd):** Many thanks to both of you, and especially to Zest Manufacturing PTY (Ltd), for all the hours and effort everyone put into the joint venture of building and refining the NAMS unit. Thank you also for your vision and willingness to get on board and support innovation and new ideas, and also contribute towards the enhancement and development of exercise- and sport science research.

**Prof Tim Noakes:** Thanks Tim for instilling a passion in all of us to question everything we read, see and do. Thank you also for following your heart and developing such a phenomenal research unit. It has been an honour to be part of this department and a privilege to work under your leadership. You are an inspiration to us all!

**Prof Vicki Lambert:** Thanks Vicki for taking a chance on me. If it was not for the opportunity you provided me, I would not have been able to continue with my studies and take the next step. Many thanks.

**Cuan van Aswegen:** Thanks for all the technical advice and support in the initial development and building of all the electronic- and force plate subcomponents.

**Jaco van Onselen and Catherine Harrison-Smith:** Thank you for all your help and many hours spent with the early phases of pilot testing and development of testing procedures.

**Trevor Cloete:** For helping me confirm and refine all the formulae for the calculation of musculotendinous stiffness and its subcomponents.

**Dr Michele Boddington:** For your great friendship and support throughout the journey, and for all the insanities we shared. They say the mind is the first to go... thank goodness it wasn't only mine!

**My fellow students in crime over the years - Yolande Harley, Dale Hudson, Delphine Oliver, Sacha West, George Mokone, Miriam Waldeck, Mark Kirkman, Jean "Ouboet" Kieser, and Hitesh "The Skinnyman" Harribhai:** Everyone needs to know they are not alone. Who better to "suffer" with than the best friends one could ask for? Thanks for everything guys...

**Prof Mike Lambert:** A special acknowledgement and many thanks to you Mike for all your support, honesty, integrity, endless patience and visionary insight. I am sure if there were ever a gold standard for supervisors, you would be it! Thanks for your financial support, friendship and backing throughout the evolution of this thesis.

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## **LIST OF ABBREVIATIONS**

**AP** = Average power  
**AT** = Achilles tendon  
**ATF** = Achilles tendon force  
**CM** = With countermovement  
**CMJ** = Countermovement jumps  
**DJ** = Drop jumps  
**EMG** = Electromyography  
**F: EMG** = Force-to-EMG  
**FLT** = Flight time  
**FT** = Fast twitch fibres  
**IEMG** = Integrated EMG  
**LOA** = Limits of agreement  
**M1** = Monosynaptic reflex component  
**M2** = Involuntary polysynaptic reflex component  
**M3** = Voluntary polysynaptic reflex component  
**MTC** = Muscle-tendon complex  
**MVC** = Maximal voluntary contraction  
**MVV** = Maximal vertical velocity  
**NAMS** = Neuromuscular and musculotendinous stiffness  
**NR** = No relationship  
**PC** = Pure concentric  
**PCBP** = Pure concentric bench press  
**PCBT** = Pure concentric bench throws  
**PP** = Peak power  
**RBP** = Rebound bench press  
**RBT** = Rebound bench throws  
**RFD** = Rate of force development  
**SJ** = Squat jumps or static jumps  
**SME** = Supra-maximal eccentric loading  
**SSC** = Stretch shortening cycle  
**ST** = Slow twitch fibres  
**WC<sup>+</sup>** = Weak positive correlation

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# **ABSTRACT:**

## **BACKGROUND:**

The stretch shortening cycle muscle action is an important part of dynamic muscle function, and enhances muscle performance by improving the force output, power output and mechanical efficiency of the working muscles when compared to isolated muscle function. Although many factors contribute towards the stretch shortening cycle muscle action, tendon elasticity seems to have a predominant role in determining the stretch shortening cycle potentiation of muscle performance.

Musculotendinous stiffness or elasticity is difficult to measure *in vivo*. Therefore, various procedures have been used in an attempt to quantify the contribution of the elastic properties of muscle and tendon to stretch shortening cycle performance. The results are variable, perhaps as a result of the different techniques utilised. Although several studies have suggested an association between a more compliant muscle-tendon complex, and enhanced stretch shortening cycle performance, this interpretation is not conclusive and needs further testing with particular attention being focused on the non-invasive, *in vivo* measurement of musculotendinous stiffness.

Accordingly, the primary goal of this dissertation was to identify the relationship between the mechanical characteristics of the muscle-tendon complex, in particular tendon stiffness, and stretch shortening cycle muscle function.

To fulfil this goal, the main aims of this thesis were:

- (i) To design and develop a testing procedure/equipment that could measure stretch shortening cycle muscle function and musculotendinous stiffness in both upper and lower body musculature
- (ii) To determine if the procedures for measuring musculotendinous stiffness in both upper and lower body musculature were repeatable.
- (iii) To determine if the procedures for measuring stretch shortening cycle muscle function in both upper and lower body musculature were repeatable.



- (iv) To refine the testing procedures, based on the experience gained in (ii) and (iii).
- (v) To determine the relationship between musculotendinous stiffness and stretch shortening cycle muscle function in both upper and lower body musculature.

## **METHODS:**

- (i) After a review of research on stretch shortening cycle testing and measurement of musculotendinous stiffness, an idea was formulated and design sketches drawn for the development of the NAMS (*Neuromuscular And Musculotendinous Stiffness*) unit. A local gym equipment manufacturer (Zest Manufacturing PTY (Ltd), Cape Town) agreed to sponsor the construction of the structural equipment. Although based on principles from previous research, the equipment was a unique design with its own features and technology.
- (ii) Musculotendinous stiffness was measured using the oscillation technique, which utilises the force-oscillations following the perturbation of a loaded muscle-group in a semi-static position. The repeatability of the test was measured over 3 days in the lower body ( $n = 10$  subjects) and upper body ( $n = 9$  subjects). Subjects were tested at relative loads of 30, 40, 50, 60 and 70 % of their maximum isometric leg press and maximum rebound bench press in the lower body and upper body studies respectively. Maximal stiffness was extrapolated from the submaximal stiffness-load relationship using a downward exponential association equation.
- (iii) The stretch shortening cycle testing procedures on the NAMS unit utilised squat jumps and countermovement jumps, and pure concentric bench throws and rebound bench throws to differentiate muscle performance potentiation as a result of the stretch shortening cycle muscle action in the lower and upper body respectively. Force- and EMG data were also collected to differentiate muscle performance in both the upper body and lower body studies. The repeatability of these tests was measured over 3 days in the lower body ( $n = 10$  subjects) and in the upper body ( $n = 10$  subjects).
- (iv) Based on the experience gained in the repeatability studies, the oscillation technique used in the lower body and upper body studies was modified to utilise an absolute rather than a relative loading protocol, and the data were fitted with a Boltzmann sigmoid equation, rather than the downward exponential association

equation to derive maximal stiffness. Minor, yet significant changes were also made to the protocols for measuring stretch shortening cycle ability in the lower and upper body.

- (v) Male subjects were tested using the modified oscillation test for the lower body ( $n = 17$  subjects) and upper body ( $n = 19$  subjects), and the submaximal and maximal musculotendinous (tendon) stiffness data were compared to the stretch shortening cycle potentiation of muscle performance using the modified techniques.

## **RESULTS:**

- (i) The NAMS unit was completed and subjected to procedural and pilot testing in July 2001, (Chapter 2).
- (ii) The measures of musculotendinous stiffness of the lower body musculature showed intra-class correlation coefficients and coefficients of variation ranging from  $R = 0.97 - 0.99$  and  $CV = 3.4 - 8.0 \%$  respectively over the range of loads tested. The average maximal stiffness of the total group was  $30.7 \pm 5.9 \text{ kN.m}^{-1}$  (Chapter 3). The measures of musculotendinous stiffness of the upper body musculature showed intra-class correlation coefficients and coefficients of variation ranging from  $R = 0.81 - 0.94$  and  $CV = 4.8 - 10.1 \%$  respectively over the range of loads tested. The average maximal stiffness of the total group was  $14.0 \pm 2.0 \text{ kN.m}^{-1}$  (Chapter 3).
- (iii) The squat jumps had an intra-class correlation coefficient of  $R = 0.98$  (95% CI: 0.94 - 0.99) and coefficient of variation (CV) of 3.6 (95% CI: 2.7 - 5.3)%. The countermovement jumps had an intra-class correlation coefficient of  $R = 0.96$  (95% CI: 0.89 - 0.99) and coefficient of variation of 6.4 (95% CI: 4.8 - 9.5)%. The acceptable force- and EMG data had intra-class correlation coefficient ranges of  $R = 0.86 - 0.99$  and  $R = 0.82 - 0.90$ , and coefficients of variation ranges of  $CV = 3.3 - 11.7 \%$  and  $CV = 16.9 - 24.0 \%$  respectively (Chapter 5). The pure concentric bench throws had an intra-class correlation coefficient of  $R = 0.99$  (95% CI: 0.97 - 1.00) and coefficient of variation of 5.8 (95% CI: 4.4 - 8.6) %. The rebound bench throws had an intra-class correlation coefficient of  $R = 0.98$  (95% CI: 0.94 - 0.99) and coefficient of variation of 7.7 (95% CI: 5.8 - 11.4) %. The acceptable force- and EMG data had intra-class correlation coefficient ranges of  $R = 0.94 - 0.96$  and  $R = 0.92 - 0.96$ , and

coefficients of variation ranges of  $CV = 3.3 - 5.4 \%$  and  $CV = 10.8 - 21.7 \%$  respectively (Chapter 5).

- (iv) The graph of stiffness vs. load fitted by the Boltzmann equation followed a less rigid path and gave a more accurate representation of the functional properties expected of a muscle-tendon complex under loading conditions than the original downward exponential association.
- (v) In Chapter 6, 17 male subjects were recruited and underwent tests to measure maximal musculotendinous (tendon) stiffness values in their lower body musculature ( $29.4 \pm 6.0 \text{ kN.m}^{-1}$ ; range  $20.5 - 39.6 \text{ kN.m}^{-1}$ ) and stretch shortening cycle potentiation. This study showed a significant ( $p < 0.0000001$ )  $12 \pm 5\%$  increase in countermovement jump height ( $28.0 \pm 3.0 \text{ cm}$ ) over squat jump height ( $25.1 \pm 2.9 \text{ cm}$ ). In this study average force was  $32 \pm 24\%$  higher in countermovement jumps than in squat jumps. Similar, yet more pronounced improvements were noted in average and peak power; average power increased by  $103 \pm 39\%$  and peak power improved by  $61 \pm 20\%$ . The major finding in this study was that there was no relationship between both maximal (tendon) and submaximal musculotendinous stiffness measures, and any of the measurements of potentiation associated with the stretch shortening cycle. In Chapter 7, 19 male subjects were recruited for the measurement of maximal (tendon) stiffness data ( $15.6 \pm 7.3 \text{ kN.m}^{-1}$ ; range  $8.3 - 35.2 \text{ kN.m}^{-1}$ ) and stretch shortening cycle potentiation in their upper body musculature. In this study average force was  $71 \pm 32\%$  higher in rebound bench throws than in pure concentric bench throws. As with the lower body data, similar, yet more pronounced improvements were noted in average and peak power; average power increased by  $157 \pm 50\%$  and peak power improved by  $121 \pm 54\%$ . Musculotendinous stiffness had a weak inverse relationship with impulse potentiation ( $r = -0.57$ ) and showed no relationship to any of the other measures of stretch shortening cycle potentiation.

## **CONCLUSIONS:**

- (i) The NAMS unit and testing procedures are reliable and the data are an accurate representation of muscle function.
- (ii) The oscillation technique was shown to be a reliable measure of musculotendinous stiffness in both lower and upper body studies. However, the determination of

musculotendinous stiffness in this format was influenced by the load against which the muscle was contracting, and the subject's body mass in the muscles of the lower body, and to a lesser extent in the muscles of the upper body. The technique needed to be refined to address these, and some additional procedural concerns.

- (iii) The majority of the vertical jump and bench throw measures using the NAMS unit were repeatable and had sufficient precision to detect changes with about 10 subjects. However, one should be cautious in the interpretation of all stretch shortening cycle potentiation measures. The mathematical manipulation of very repeatable data sets e.g. squat jump height and countermovement jump height in the lower body, and pure concentric bench throw height and rebound bench throw height in the upper body, seems to be flawed when combined and expressed as differences, ratios or relative change indices, as this reduces the reliability and increases the variability of these derived measures. However, to represent the potentiation in muscle performance as a result of the stretch shortening cycle muscle action, one needs to relate the pure concentric measures with the stretch shortening cycle measures, and at present there does not seem to be a better alternative way to quantify this relationship. It is important, in future research, to acknowledge and understand the limitations of the derived measures of potentiation expressed in this way.
- (iv) The oscillation technique was modified to an absolute rather than a relative loading protocol, and the formulation for deriving maximal stiffness was changed from the downward exponential association equation to the Boltzmann sigmoid equation. This seemed to be a more accurate way of predicting musculotendinous stiffness, and the stiffness-load relationship, than the original method. The modifications in the stretch shortening cycle testing procedures of the lower and upper body reduced the experimental error involved and made the testing procedures more accurate in their representation of the data.
- (v) In Chapter 6, tendon elasticity (stiffness) was measured in the lower body. Next, tendon elasticity was related to the associated stretch shortening cycle performance. It was concluded that tendon elasticity, as an isolated component of muscle function, was not related to the associated stretch shortening cycle muscle performance in the lower body. In Chapter 7, tendon elasticity (stiffness) was measured in the upper body and related to the associated stretch shortening cycle performance of the upper body musculature. It was concluded that tendon

elasticity, as an isolated component of muscle function, had a weak relationship to stretch shortening cycle performance in the upper body. On the basis of this weak relationship, it can be argued that the isolated measurement of tendon elasticity has little value in explaining stretch shortening cycle performance and the mechanisms involved in muscle performance potentiation resulting from stretch shortening cycle muscle actions. As with the lower body, further research in this area should focus on the integration of, and combined interaction between, the numerous mechanisms involved in the stretch shortening cycle action.

## **SUMMARY OF FINDINGS:**

The potentiation of muscle performance during stretch shortening cycle activities occurs regardless of different muscle and tendon elastic characteristics. Based on these data, it is proposed that the neuromuscular and elastic properties of the muscle-tendon complex may contribute in varying proportions towards the potentiation in muscle performance, associated with the stretch shortening cycle muscle action.

## **CHAPTER 1:**

# **Literature Review on tendon stiffness and stretch shortening cycle performance**

## **INTRODUCTION AND SCOPE OF THESIS**

This project was initiated with the primary aim of developing a repeatable technique of testing for stretch shortening cycle ability. In pilot studies in our unit, it was found that vertical jumping techniques were not repeatable and hence lacked the required precision for mechanistic studies on muscle function. The main problem with the technique was that the subjects did not always jump and land in the same place. This initiated an extensive literature review on stretch shortening cycle testing procedures to gain insight into another approach to this problem.

During the literature search, some studies generated ideas about modifying a Smith machine to measure stretch shortening cycle function, in both upper (42;119;153-156) and lower body musculature (72;74;150;154). An advantage of this equipment was that it limited movement in one plane, and therefore extraneous movement and displacement would also be limited. This would have the potential to improve the repeatability of the measurement of stretch shortening cycle activity using vertical jumping and bench throws for the lower body and upper body musculature respectively.

Another interesting finding in the review was that numerous papers suggested that tendon elasticity had a significant role to play in stretch shortening cycle muscle function, and also that tendon elasticity, might be a determining factor in stretch shortening cycle improvement of muscular performance (97;101;133;136;151-153;156). This raised the question of how one could accurately measure musculotendinous stiffness, in particular tendon stiffness or –elasticity, to gain a better understanding of muscle function during the stretch shortening cycle.

Therefore, it was decided to design and build a modified Smith machine and force plate, together with additional attachments, which later collectively became known as the NAMS unit (**N**euromuscular **A**nd **M**usculotendinous Stiffness unit) for the testing and measurement of musculotendinous stiffness and stretch shortening cycle performance. Additionally, the testing procedures for stretch shortening cycle muscle function and musculotendinous stiffness needed to be developed, and subjected to pilot testing.

The scope of this thesis therefore was a comprehensive review of the literature on stretch shortening cycle and musculotendinous stiffness (Chapter 1). The experimental part of the thesis began with the design and development of the NAMS unit and testing procedures described in Chapter 2. This was described in detail, as it formed the foundation of the experimental phase of the thesis. After the design, building and pilot testing were completed, a formal repeatability analysis of the developed techniques of muscle function was performed.

Chapter 3 includes repeatability studies on the measurement of musculotendinous stiffness in both the lower and upper body musculature. Certain questions were raised about the technique of musculotendinous stiffness measurement in Chapter 3. Chapter 4 addressed some of these methodological concerns and refined the technique. Chapter 5 comprises the repeatability studies on the measurement of stretch shortening cycle muscle function in both the lower and upper body musculature.

Chapter 6 and Chapter 7 examined the relationship between musculotendinous stiffness and stretch shortening cycle performance in the lower body and upper body respectively.

Chapter 8 provides a summary of the conclusions and theories regarding the stretch shortening cycle, and how musculotendinous stiffness relates to this component of dynamic muscle function.

## **DEFINITION OF TERMS**

There has recently been an extensive debate on the correct nomenclature that most accurately describes muscle function (49). Three main types of muscle contractions namely concentric-, isometric- and eccentric contractions have generally been used to

describe muscle function during movement (91). The main terminologies under question are “**concentric**”, “**eccentric**” and “**contractions**” (49). Other names to describe these respective movements have been “**shortening contractions**” and “**lengthening contractions**” (49). The term concentric/shortening and eccentric/lengthening “**muscle actions**” as opposed to “**contractions**” has also been used (49). Even though the ability of the present terminology to correctly describe these different forms of muscle function has been queried (49), the action of the muscle and tendon during these movements remains the same. Therefore, even though it is a controversial and contentious topic at present, for the purpose of this review the terms “**eccentric contraction**”, “**isometric contraction**” and “**concentric contraction**”, as they have frequently been described, have been maintained to describe isolated muscle function.

It is common knowledge that muscles can either decelerate or accelerate the body or body part (33). Concentric contractions occur when the muscle length shortens during contraction and the net muscle moment is in the same direction as the change in joint angle, resulting in acceleration of the body or body part (91). Isometric contractions occur when joint angles remain constant and no visible movement occurs. A consequence of an isometric contraction is that the body or body part is fixated or stabilised (91). When the force applied to the muscle is greater than the force which the muscle generates, the muscle will lengthen (107). Eccentric contractions occur when the muscle lengthens while contracting (93), and the net muscle moment is in the opposite direction to the change in joint angle (91). The muscles are actively recruited (93), and their function is to decelerate the body or body part (91).

To develop specific training techniques and improve athletic performance, one needs to understand the specific mechanisms underlying dynamic muscle function (150). Muscle function during exercise or normal movement seldom involves pure forms of isolated eccentric, concentric or isometric contractions (93;95). Instead, movement is often characterised by a combination of these actions, particularly in situations involving impact or gravitational loading (93). During this situation there is cycling between eccentric and concentric contractions, with the concentric being preceded by eccentric action (36;91;92;95).



During most sporting activities, the muscles are primed by a countermovement in the opposite direction, before the movement is initiated (18;22;42;137). A countermovement is a period of force reduction and deceleration immediately preceding an intended movement (159). The active muscles first have to absorb the momentum by eccentric contraction and then immediately follow it by a concentric contraction in the opposite, intended direction (132). This cyclic relationship is referred to as the “**stretch shortening cycle**” (SSC), also known as plyometric muscle action (17;22). In other words, the stretch shortening cycle describes muscle function where an eccentric contraction is immediately followed by a concentric contraction (30;64;100;101;136-138). This stretch shortening cycle forms a natural component of dynamic muscle function (50;103;139).

The stretch shortening cycle demonstrates significant improvements in muscle performance compared to isolated muscle function. This enhancement in performance has frequently been referred to as “**potentiation**” e.g. myoelectric “**potentiation**” or jump height “**potentiation**”. For the remainder of this review, the term “**potentiation**” will represent the enhancement or improvement in the relevant performance parameter.

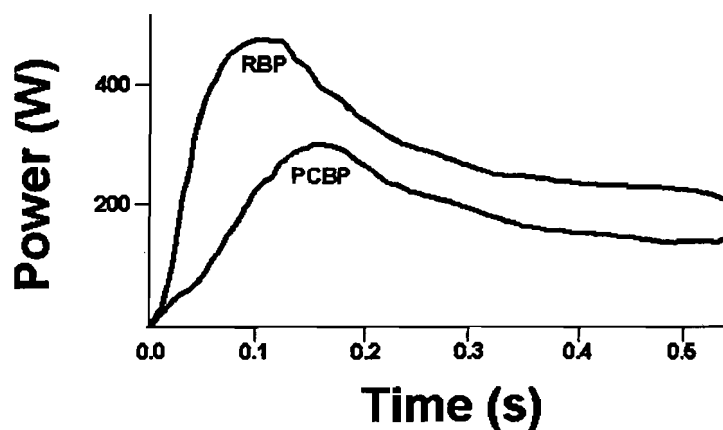
## **STRETCH SHORTENING CYCLE**

### **INTRODUCTION**

The stretch shortening cycle results in increased muscle performance when compared to isolated concentric contractions performed without prior stretch (26;27;42;47;56;75;93;101;136-140). This is predominantly visible in an increased power output and improved mechanical efficiency (107). Hubley and Wells (78) demonstrated that countermovement (stretch shortening cycle) jumps performed greater work compared to static or squat (concentric) jumps. Pre-stretching active muscle increases the concentric work done, but more specifically improves the power output of the involved muscles, by reducing the time over which the work is performed (33). The major effect of eccentric pre-stretch on concentric performance potentiation therefore seems to rather be a greater power output than a greater amount of work performed (33).

In essence, the eccentric phase of the stretch shortening cycle positively influences the subsequent concentric contraction, making it more powerful than an isolated concentric

contraction (32;91;92;119;137;138). This power potentiation effect can clearly be seen when the force-velocity curve (95) and power-time curve of a stretch shortening cycle muscle action is compared with an isolated concentric contraction (156). In Figure 1, the power output in the concentric contraction in the rebound bench press, which utilises the stretch shortening cycle, is noticeably larger than the pure concentric bench press, which is an isolated form of concentric contraction. Walshe et al. (150) also found that when a squat was preceded by an active stretch, more work was performed over the initial 300 ms of the concentric movement. This finding seems to support the abovementioned results of Wilson et al. (156).



**Figure 1:** The mean instantaneous power output for the five most compliant subjects for the first 0.5 seconds of the concentric phase of the rebound bench press (RBP) and pure concentric bench press (PCBP). Modified from Wilson et al. (156).

Following the eccentric pre-stretch during a stretch shortening cycle movement, the force is already high at the beginning of the concentric movement (28). This leads to an increased acceleration of the mass to which the muscle is attached and thus also contributes to a greater power output (33). Bosco et al. (30) showed a 41% increase in the force generated, and a 43% increase in power and higher jumps heights, when comparing countermovement jumps to squat jumps, which supports the abovementioned conclusion. This cyclic action of muscle function is a prerequisite for success in sports with explosive movements such as gymnastics, basketball, rugby and athletics to name but a few (149). Several training principles and treatments in sport rehabilitation and physiotherapy also

utilise the principle that by eccentrically pre-stretching a muscle, its ensuing concentric contraction will be enhanced (139). This supports the importance and value of testing for the stretch shortening cycle and the effects of various training strategies, specific activities, fatigue and preceding muscle actions in the development or attenuation of stretch shortening cycle muscle function.

## ***ISOLATED MUSCLE FUNCTION***

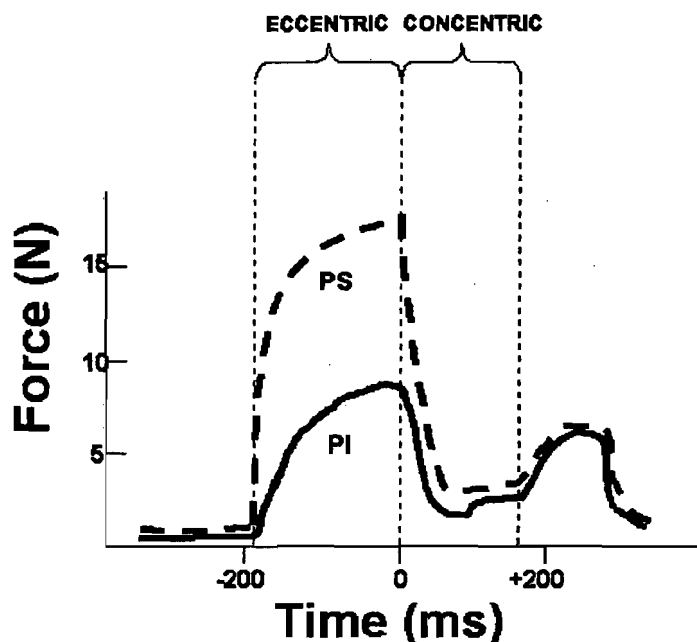
Force and speed are influenced by the structure, mechanics and elastic characteristics of muscle (91). According to the well-characterized relationship between muscle force and the speed of contraction it is known that with increasing speed of shortening, muscle force output decreases exponentially (1;56;66;126;127). However, this is only true with concentric muscle contractions. During eccentric contractions, with increasing velocity of muscle lengthening, force output increases in a non-linear hyperbolic manner (1;127). An initial increase in force output up to a movement speed of about  $90^{\circ} \cdot s^{-1}$  followed by a gradual decrease (68) or maintenance (66) in force output of the muscle has also been shown. The maximum force produced during eccentric contraction is significantly higher than the maximum force during a concentric contraction, and may even be twice as high as the maximum force generated during isometric contractions (1;34;95;124).

Dietz et al. (41) clarify that a contracted muscle is capable of developing additional tension or force when it is forcibly stretched. Rack and Westbury (126) attribute this additional force to the distortion of the actin-myosin cross-bridges within the myofibrils.

## ***COMBINED MUSCLE FUNCTION***

Based on the abovementioned force-velocity relationships, the increased force output of the muscles during a high velocity eccentric contraction, followed by a reduced force output in a high velocity concentric contraction, increases the range of force applied to the muscle structures (66). The range of force can be defined as the change in force, or the difference between the maximum force generated during the eccentric contraction and the maximum force generated during the subsequent concentric contraction at measured peak velocity of contraction.

In reality this increase in range of force can only occur by pre-stretching the muscle via eccentric contraction and following it with concentric contraction, as occurs during stretch shortening cycle movements. This change in force is also significantly more pronounced in faster as opposed to slower movement speeds (126). During eccentric contraction, force increases above maximal isometric force (1;33), and decreases rapidly into concentric contraction, thereby increasing the range and thus change in force (79) (Figure 2). The high eccentric force also leads to a significantly higher initial force of concentric contraction (79;139), which assists in acceleration of the mass or body (33). One should also note that due to the elastic ability of the muscle-tendon complex (MTC), the decrease in force in the first part of the concentric phase of a stretch shortening cycle movement, is also more pronounced (79).



**Figure 2:** The force generated by pre-stretch (PS) is considerably higher than the force generated by pre-isometric (PI) contractions. The relative decrease in force during the ensuing concentric contraction is greater for PS than PI conditions. Modified from Huijing (79).

The range and decrease in force as discussed above, has an important role in stretch shortening cycle energy dynamics (151).

## **ENERGY TRANSFER**

During concentric work the muscle derives energy from two sources (i) chemical energy and (ii) mechanical energy. The chemical energy results from the metabolism of various substrates, and the elastic properties of the tissues supply mechanical energy. Many of the findings concerning energy dynamics are made on the basis that muscles are assumed to have elastic properties (33).

Arampatzis et al. (5) indicate that there is an energy trade-off between muscles, tendons and ligaments during human movement. Coveney (39) proposes that the muscle-tendon complex, because of its capacity to generate, absorb and recover elastic energy, plays a major role in efficient and controlled movement. According to Alexander and Bennet-Clark (2), muscles, tendons and their skeletal support structures all have elastic properties. These properties play an integral part in the mechanical functioning of intact skeletal muscle (25). Bosco et al. (25) suggest that the right shift in the force-velocity curve (i.e. increased force at high velocities), when utilising the stretch shortening cycle compared to pure concentric contraction, reflects the storage and utilisation of elastic energy by the involved muscles. They also state that the involved muscles function like a spring, and that elastic energy restitution plays an important role in the dynamics of this type of muscle action.

## **ECCENTRIC FORCE**

Energy storage occurs mainly during the eccentric deceleration phase (24). Eccentric contraction stores a significant amount of potential energy in the muscle-tendon complex when the series elastic components are stretched (36;139), and subsequent concentric contraction can partly recover this energy (33;74;89;134). However, the actual amount of elastic energy stored when muscles are stretched against a load is unknown (32).

With most movements of an eccentric nature, the muscles contract to brake or decelerate the movement. The active muscles are forcibly stretched during this phase (12;60) and the resultant high forces which are produced, favour the storage of elastic strain- or potential energy in the muscle-tendon complex (17;20;33;36;92;93;131). Essentially, when a force stretches a spring, in this case the muscle-tendon complex, the work done by

the force during stretching is stored as elastic strain energy (132). The rapid decrease in the force applied to the muscle-tendon structure during concentric contraction as mentioned earlier, liberates the mechanical- or elastic energy (116) stored during eccentric contraction (79;136). In most movements, especially movements involving relatively high power outputs, the muscles undergo the stretch shortening cycle by contracting concentrically immediately following eccentric pre-stretch, thereby utilising the stored elastic energy.

The amount of stretch of the tendon and therefore energy stored depends largely on the force exerted upon the muscle-tendon complex. The greater the eccentric force applied, the greater the muscle-tendon complex extension, and thus storage of potential- or elastic energy (20;33). The amount of energy stored in the series elastic elements i.e. tendon, aponeurosis and actin-myosin cross-bridges, relates to the force developed by the involved muscles (17;18). Hooke's law states that the higher the force that stretches a spring, in this example the muscle-tendon complex, the greater the release of energy will be (17;64). Also, the higher the force, the greater its resultant shortening after release, and because of this, it correspondingly liberates more elastic energy (34). The reasoning therefore follows that as a consequence of the increased force generated during eccentric stretching of the muscle-tendon complex, there will be an increased storage (20;33;141), and consequently an increased release of elastic energy during the concentric portion of a stretch shortening cycle movement (34;64;141). This storage and utilisation of mechanical- or elastic energy therefore depends largely on the occurrence of the stretch shortening cycle.

#### **ECCENTRIC-CONCENTRIC TRANSITION**

The phenomenon of increased concentric performance after active stretch has, however, been shown to be transient of nature (34;35). If this eccentric-concentric transition is performed quickly, the release of mechanical energy will contribute appreciably to the augmentation in concentric muscle performance (25). Part of this stored energy is recovered and utilised during the subsequent concentric phase of the stretch shortening cycle (18;26;74;93), and part of the stored energy is lost as heat (7). Alexander (1) showed that the elastic recoil of tendons return approximately 93% of the energy stored during eccentric stretching and only loses about 7% of the energy in the form of heat. If,

however, the time transition between eccentric and concentric actions is too long, most of the stored elastic energy is lost in the form of heat, instead of being utilised in the following concentric contraction (91;121;144). This would cause less elastic energy assistance and thus lead to a reduction in the potentiation of muscle performance resulting from the stretch shortening cycle.

The force generated at the end of the eccentric stretch is dependent on both the amplitude and velocity of the stretch (64). Consequently, efficient storage and utilisation of elastic energy during stretch shortening cycle movements is reliant on the rate of stretch, force generated, and transition time between stretching and shortening of the involved muscles (27;33;137).

Svantesson and Grimby (138) and Svantesson et al. (137), substantiating the abovementioned conclusions, found that the percentage increase in concentric force with prior stretch, was significantly higher after a pre-stretch speed of  $240^{\circ} \cdot s^{-1}$  compared to  $120^{\circ} \cdot s^{-1}$ . Nicol et al. (121) studying the effects of fatigue on stretch shortening cycle function suggest that a diminished stretch shortening cycle potentiation is the result of an increase in the transition-time between eccentric and concentric contractions, which leads to a reduction in elastic energy storage and utilisation.

### CROSS-BRIDGES

Shorten (132) suggests that elastic energy storage is due to the nature of cross-bridge turnover and stress-relaxation of the tendon and the muscle-tendon complex, and that elastic recoil can only contribute to potentiation of performance if concentric action immediately follows eccentric stretch. The duration of binding of an actin-myosin cross-bridge is limited, so a short transition between stretch and shortening is advantageous to gain maximal benefit from pre-stretch (54), and therefore utilisation of stored series elastic energy in the concentric or shortening phase (91). Confirmation of this was found in the fact that when muscle in a tetanic state of contraction is stretched using a "**ramp-and-hold**" type of protocol, the tension or stiffness initially rises during the active stretch, and subsequently dissipates at the beginning of the hold phase (64). Elastic energy storage can only occur while the cross-bridges are attached. When the cross-bridges detach,

through either relaxation or over-stretching, the elastic strain energy is released in the form of heat (132).

Ettema and Huijing (46) however remind one that nothing comes completely free of metabolic cost; everything costs energy. Muscles use chemical energy to perform mechanical work and also to store elastic energy (128). It is during eccentric stretching that the elastic energy, which is later released in the ensuing concentric contraction, is stored in the series elastic elements (137). During the eccentric phase of stretch shortening cycle movements, the muscle increases its tension to resist the forceful stretching of the muscle-tendon complex (137), and this resistance to stretch requires metabolic energy (2;36;46). Even if no mechanical work is performed by the muscle fibres, metabolic energy is required to exert force (1).

It has nonetheless been shown that recovery of elastic energy significantly compensates for the additional energy spent on sustaining the eccentric contraction during active pre-stretch of the involved muscles (33).

## **SUMMARY OF BACKGROUND TO THE STRETCH SHORTENING CYCLE**

The stretch shortening cycle of muscle function utilises a combination of the three main contraction types, eccentric-, isometric- and concentric contractions, to optimise muscle performance during dynamic movement. Performing these contractions sequentially, using the stretch shortening cycle enhances muscle performance compared to muscle performance during isolated concentric contraction, i.e. the stretch shortening cycle increases both the force- and power output of the concentric phase, thereby making it more efficient. The stretch shortening cycle muscle action, however, needs to be performed rapidly, with a short transition-time between eccentric and concentric contractions, to gain maximum benefit from the altered force dynamics and the storage and utilisation of elastic energy.



## **MEASUREMENT OF STRETCH SHORTENING CYCLE**

The measurement of stretch shortening cycle ability has generally used variations of vertical jumping techniques (13;73;89;95;135;149). This is possibly due to vertical jumping being a functional movement and a relatively easy technique to control and measure. In human studies, the effects of the stretch shortening cycle on performance have mostly been tested using vertical jumps performed with and without countermovement (132) to vary the eccentric pre-loading conditions.

Stretch shortening ability in humans has also been tested using an isokinetic dynamometer (137-140), an upper body sledge apparatus (58;59) and bench press and bench throw variations (42;119;153;156).

### **VERTICAL JUMP**

Peak performance in, and comparison between, vertical jumping types is usually measured by relationships between:

- Vertical take-off velocities ( $V_v = \frac{1}{2} \times t_{air} \times g$ ) (7;24-27;29;30;89;97;101),
- Net impulses ( $NI = m \times V_v$ ) (24;26),
- Power outputs ( $(F \times s) / t$ ),
- Height of rise of the center of gravity ( $h = V_v^2 / 2g$ ) (24;26;89;97;101) (or  $h = 1.226 \times (t_{air})^2$ ) (7),
- Average force outputs ( $F = NI / t$ ) (24-27;29;30),
- Vertical ground reaction force (VGRF) (24;26;69;105)
- And various other force-time measures, such as instantaneous power ( $P = V_v \times VGRF$ ) and maximal explosive power (Peak W) (69;159).

Where:

- $V_v$  = Vertical velocity ( $m.s^{-1}$ )
- $t_{air}$  = Flight time (s)
- $g$  = The acceleration of gravity ( $9.81 m.s^{-2}$ )
- $m$  = Mass (kg)
- $F$  = Force (N)
- $NI$  = Impulse (Ns)

**s** = Distance (m)  
**t** = Time (s)  
**h** = Height (m)  
**P** = Power (W)

These tests are variations of jumps used in conjunction with various force platforms and force plates e.g. Kistler, AMTI, and Bertec.

The tests have ranged from the standard type of vertical jumping tests, i.e. squat jumps (SJ); countermovement jumps (CMJ); and drop jumps (DJ) from various heights (7;13;21;24;26;62;73;89;144;149;151), to the same jump variations with different types of angled sledge apparatus (9;12;13;83;92;104;135;159).

Vertical jumps are multi-joint movements and are a functional expression of the combined muscle action of all the muscles involved (159). Most studies using vertical jump differentiate the performance during squat jumps, countermovement jumps, and drop jumps.

A **squat jump** is a vertical jump performed from a stationary 90° knee angle position with no preparatory movement allowed and consists of an isolated concentric contraction. The stationary bottom position should also be maintained for ~2 seconds before initiation of the jump to limit any stretching effects (6;26;30;89;97;158).

Using a modified sledge apparatus, Zamparo et al. (159) compared squat jumps at knee angles of 70, 90, 110, 130 and 150° performed on a force-platform with the exact same jumps on an inclined sledge apparatus. They found that the power outputs in the sledge jumps were significantly lower than those in the standard squat jump comparisons.

With the standard squat jump test, it is not possible to completely eliminate all countermovement, whereas with the sledge apparatus (Multipurpose Ergo dynamometer) it was (159). Zamparo et al. (159) therefore attributed this difference in performance to the recovery of additional elastic energy in the vertical jumps, which was not possible on the sledge apparatus. This has important methodological implications for future reference.

A **countermovement jump** is a vertical jump starting from an erect standing position with a countermovement preceding the jump; in other words an eccentric contraction preceding the concentric contraction (6;25;26;30;89;97).

A **drop jump** is a drop from an elevated position and a pre-determined height followed immediately on landing by a subsequent maximal double-leg vertical jump (6;7;17;26;42;89;134), hence also incorporating the stretch shortening cycle under various eccentric loading conditions.

The majority of studies use flight times and vertical ground reaction force to calculate vertical jump height. Using ground reaction force and flight time calculations during vertical jumps, it is however important for the accuracy of the measurement that the subject's body position is the same at landing as it was at take-off (158). For this reason subjects are instructed to land in a fully extended position before flexing hips, knees and ankles to reduce and absorb impact (158).

This form of testing generally focuses on the augmentation in concentric muscle performance, improvement in jump height and enhancement of force-time kinematics as indicators of stretch shortening cycle ability (Table 1).

**Table 1:** Evidence of stretch shortening cycle potentiation of muscle performance in the lower body musculature.

Stretch shortening cycle potentiation of performance in the lower body				
Reference	Variable	Test	Measured	Potentiation
Walshe et al. (151) Males (n = 23)	Jump height	SJ CMJ	32 ± 5 cm 36 ± 5 cm	14 ± 6 %
Komi and Bosco (89) Female Phys. Ed. Students (n = 25)	Jump height	SJ CMJ	19 ± 4 cm 23 ± 4 cm	21 %
Komi and Bosco (89) Male Phys. Ed. Students (n = 16)	Jump height	SJ CMJ	36 ± 5 cm 40 ± 7 cm	14 %
Komi and Bosco (89) Male volleyball players (n = 16)	Jump height	SJ CMJ	37 ± 4 cm 43 ± 5 cm	17 %
Bosco and Komi (26) Females (n=113)	Jump height	SJ and CMJ	-	12-23 %
Bosco and Komi (26) Males (n=113)	Jump height	SJ and CMJ	-	10-20 %
Bosco and Komi (24) Male Phys. Ed. Students (n = 34)	Jump height	SJ CMJ	36 ± 5 cm 42 ± 6 cm	-
Häkkinen et al. (72) Powerlifters (n = 4)	Jump height	SJ CMJ	38 ± 7 cm 39 ± 9 cm	-
Häkkinen et al. (72) Bodybuilders (n = 7)	Jump height	SJ CMJ	40 ± 3 cm 44 ± 5 cm	-
Häkkinen et al. (72) Wrestlers (n = 3)	Jump height	SJ CMJ	37 ± 6 cm 43 ± 6 cm	-
Häkkinen et al. (74) Elite male weight lifters (n = 14)	Jump height	SJ CMJ	41 ± 4 cm 49 ± 5 cm	-
Shorten (132) Plantar flexors Subjects unknown	Jump height	SJ CMJ	4.1 cm 5.0 cm	-
Bobbert et al. (18) SJ at various positions within the ROM Male volleyball players (n = 6)	Jump height	SJ1 SJ2 SJ3 CMJ	45 ± 4 cm 46 ± 5 cm 45 ± 5 cm 48 ± 4 cm	-
Bosco et al. (27) Power athletes (n = 14)	Force	SJ and CMJ	-	423 ± 237 N (66 %)
Bosco et al. (27) Power athletes (n = 14)	Power	SJ and CMJ	-	1158 ± 546 W (81 %)
Bobbert et al. (18) SJ at various positions within the ROM Male volleyball players (n = 6)	Force	SJ1 SJ2 SJ3 CMJ	1006 ± 218 N 1187 ± 335 N 905 ± 142 N 1708 ± 336 N	-
Bosco et al. (29) Small amplitude jumps Male (n = 10) and female (n = 4) Power athletes	Force	SJ CMJ	746 ± 109 N 1308 ± 253 N	-
Bosco et al. (29) Large amplitude jumps Male (n = 10) and female (n = 4) Power athletes	Force	SJ CMJ	506 ± 164 N 752 ± 221 N	-
Bosco et al. (30) Well-trained male athletes (n = 3)	Force	SJ CMJ	486 N 684 N	41 %
	Power	SJ CMJ	1240 W 1774 W	43 %
Kawakami et al. (85) Plantar flexion exercise Males (n = 10)	Max force	PC CM	3081 ± 667 N 4055 ± 655 N	-
	Max power	PC CM	332 ± 109 W 408 ± 144 W	-
	Mean power	PC CM	236 ± 88 W 158 ± 48 W	-

**Index:** SJ = squat jumps, CMJ = countermovement jumps, PC = pure concentric,  
CM = with countermovement

### ***Energy measures in vertical jumping***

Asmussen and Bonde-Peterson (7) were the first researchers to use the technique of drop jumps (132) and measured elastic energy contributions during countermovement jumps and drop jumps from various heights. They compared the amount of energy absorbed during the downward movement (eccentric/negative work) in countermovement- and drop jumps, and compared it to the energy utilised during the subsequent upward movements (concentric/positive work). The baseline measures were calculated from the pre-tested squat jumps. The positive work in squat jumps was used as the baseline reference for any change in positive work during the following jumps

The increase in elastic energy return was calculated as the amount that positive work increased from baseline measures, expressed as a percentage of the energy stored during negative work in the downward phase of each jump. They found that utilisation of stored elastic energy increased, as the drop jump height increased, until a drop height 40 cm, whereafter it decreased (7). This reduction in performance after a drop height of 40 cm could be attributed to excessive eccentric loading with corresponding neural inhibition of concentric performance.

### ***Stretch shortening cycle ability in vertical jumping***

When performing vertical jumps incorporating the stretch shortening cycle such as countermovement jumps, subjects achieve greater jump heights than found using squat- or static jumps (18) (Table 1), suggesting that the short time-transition between eccentric and concentric contraction in countermovement jumps results in the utilisation of elastic energy (24). In the past numerous researchers have measured the stretch shortening cycle ability of the muscle-tendon complex as the difference between squat jump height and countermovement jump height expressed as a percentage change (6;73;97;101;151;152):

$$\text{Percentage Difference (Stretch shortening cycle ability)} = (\text{CMJ-SJ})/\text{SJ} \times 100\%$$

According to this interpretation, the difference in squat and countermovement jump height reflects the stretch shortening cycle ability of the individual, or the ability to utilise pre-

stretch and therefore the stretch shortening cycle (132). In other words, the difference between squat jumps and countermovement jumps describes the ability of an individual to store and utilise elastic energy (73). However, this interpretation does not consider a neural contribution to the potentiation, which would result in altered muscle recruitment patterns and may be independent of the intrinsic elastic ability of the muscle-tendon complex.

### ***EMG and muscle recruitment in vertical jumping***

To measure the various components during dynamic stretch shortening cycle actions, i.e. stretch reflex potentiation or muscle-tendon complex elasticity, a measure of EMG should be included in the testing protocol. Surface EMG can be measured as a rough estimation of neuromuscular activation (23) or activation of the motorneuron pool (41).

An important factor that needs consideration in measuring surface EMG is the concept of electromechanical delay. Electromechanical delay has been defined as the time-delay between onset of muscle activation and development of force (37;160). The cause of this delay has been attributed to: (i) muscle membrane action potential propagation, (ii) release of  $\text{Ca}^{2+}$  from the sarcoplasmic reticulum, (iii) excitation-contraction coupling and force development, and (iv) stretching of the series elastic components of muscle (28;37;160). Previous findings also suggest that muscle with a higher type II fibre composition; greater force of contraction, greater rate of force development and a stiffer series elastic component could elicit a shorter electromechanical delay (160). From a methodological perspective, when EMG is related to ground reaction forces, one has to however assume that electromechanical delay is either negligible or constant across trials, although it may differ between subjects (60).

Isear et al. (82) performed a study to measure the muscle recruitment patterns in the lower extremities during the unloaded squat movement. They measured EMG activity in the gastrocnemius, vastus medialis, vastus lateralis, rectus femoris, hamstrings and gluteus maximus muscles. They also placed a common ground electrode on the anterior proximal tibia. In this study, they excluded certain muscles from being major contributors towards the squat based on their normalised contribution towards the movement. For normalisation of EMG they also suggest that closed chain testing is biomechanically more

specific in nature, and is preferable over the use of open chain methods for testing maximum voluntary contraction (MVC) in multiple function muscles (82).

The vastii muscles work at the highest percentage of maximum during the squat movement. The vastus medialis and vastus lateralis contributed ~68% and ~63% of MVC respectively. The greatest recruitment of the vastii muscles occurred in the 90-60° arc. It was assumed that the greater activity in this arc was needed to propel the body upwards after the 90° bottom position (82).

Rectus femoris contributed about ~48% of maximum voluntary contraction (MVC) (82), with different EMG patterns than the other muscle groups (30;82). This difference may be explained because rectus femoris acts as a bi-articular or two-joint muscle and therefore has different activation patterns (30;82).

The contribution of the hamstrings, gluteus maximus and gastrocnemius muscles to the movement during the squat was very low as their activation was ~12%, ~12% and 7-11% of MVC activation respectively. During a loaded squat or vertical jumps, one could assume that the proportional contributions of the various muscle groups mentioned should remain about the same.

Hubley and Wells (78) using a slightly different methodology, showed that the muscles at the ankle, hip and knee joint contributed on average 23%, 28% and 49% respectively to the positive work done during vertical jumps.

With jumping movements such as drop jumps or submaximal hops, the kinematics and dynamics may be slightly different. Based on previous measures of EMG (12;13;21) it would be logical to assume that the calf muscles (gastrocnemius or soleus) would have an increased EMG activity, and therefore activation and contribution towards vertical movement in these jumps.

Based on the above, it would be advisable to measure EMG activity in either the vastus medialis or -lateralis muscles in standard vertical jumps and including either the gastrocnemius or soleus muscles in drop jumps or submaximal hops(7;12).

Häkkinen et al. (74) used the unloaded and loaded squat jump-, countermovement jump- and drop jump techniques from various heights to study EMG and force-production characteristics of elite weight lifters. A barbell was placed on the shoulders and 40, 80, 100, 140 and 180 kg loads were used. Unloaded drop jumps were also performed from 20, 40, 60, 80 and 100 cm heights. EMG activity of the vastus medialis and -lateralis muscles were measured. EMG activity measured during the concentric contraction phase did not differ significantly between unloaded conditions, except most importantly in the drop jump from 100 cm where EMG was significantly diminished. This can be explained, because when the muscle-tendon complex is subjected to high impact forces, the reflex activation responses are reduced by Golgi tendon organ inhibition, as a protective response of the central nervous system (58). Gollhofer et al. (62) also found a reduction in the concentric force production during drop jumps from a height of 56 cm when compared to countermovement jumps, and drop jumps from heights of 24 cm and 40 cm. These results could also be explained due to inhibition of muscle contraction resulting from the stimulation of Golgi tendon organ.

In the loaded jumps, there were greater countermovement jump heights compared to squat jump heights at all loads (74). However, there were no significant differences in average concentric EMG during the countermovement jumps. These findings indicate no myoelectrical potentiation, whereas there was a performance potentiation. This suggests a better utilisation of elastic energy in all countermovement jump conditions with increasing loads (i.e. 40 - 180 kg) (74).

### ***Neuromuscular control mechanisms involved in vertical jumping***

Vertical jumps utilising eccentric pre-stretch have the additional component of neuromuscular involvement i.e. neural preactivation and reflex potentiation (77;144), which can also contribute to enhance concentric performance.

Gollhofer et al. (62) measured the force-length relationship of the achilles tendon-triceps surae complex during the various vertical jump conditions. They concluded that because the tendon and muscle are connected in series in a muscle-tendon complex, the combined structural unit determines stiffness, and that stiffness, is controlled by neuromuscular activation. This would largely imply that stretch shortening cycle ability is attributed



primarily to neural control mechanisms. Gollhofer et al. (62) also show that a distinct stiffness of the muscle-tendon complex is needed throughout for the stretch shortening cycle of muscle function to be effective. They further propose that the tendon is the main elastic buffer during the initial phases of eccentric loading in the stretch shortening cycle, and that the amount of stretch, and therefore stiffness thereafter, would be a function of the preactivation and neural activation of the involved musculature (62).

Preactivation refers to the muscle being activated before ground contact to regulate stiffness for the storage and utilisation of elastic energy (7;13;62;93;94;121). Muscle preactivation seems to be the most sensitive mechanism contributing to drop jump performance (83), as preactivation increases significantly more than the other time-defined EMG measures with increased loading of the muscles (60). Preactivation nonetheless seems to be in part pre-programmed, thereby preparing the muscle-tendon complex for impact by increasing muscle stiffness (60), yet can further be altered or modified through proprioceptive, vestibular (113) and visual input (13). It has been suggested that preactivation is pre-programmed and controlled either by the higher brain centres, or is centrally mediated (12;77) and furthermore, can be developed by previous learning experiences.

Gollhofer and Kyröläinen (60) demonstrated that anticipation plays an important role in preactivation and that this in turn affects the stretch shortening cycle enhancement in drop jumps. They showed that by artificially reducing body weight on impact during a drop jump, the preactivation EMG was considerably less compared to a full body weight drop jump from the same height, and did not provide the muscle-tendon complex with sufficient stiffness to improve performance. Dietz et al. (41) agree that the early increase in EMG might be caused by central programming gauged by previous experience, combined with visual anticipation. They further suggest, however, that a large proportion of the EMG response after impact is reactive rather than pre-programmed. Melville-Jones and Watt (113) on the other hand, inferred from their results, that pre-programmed neuromuscular activity rather than stretch reflex activation causes muscular deceleration on landing.

Avela et al. (13) measured the interaction between preactivation, measured by surface EMG, and muscle output of the triceps surae and vastus lateralis muscles during stretch shortening cycle loading. They found that preactivation, started before ground contact

and that the amount of EMG, and therefore amount of preactivation, increased linearly with increasing eccentric load. On the basis of their findings Avela et al. (13) suggest a strong link between preactivation, reflex facilitated muscle stiffness and utilisation of stored elastic energy in the concentric push-off phase.

Gollhofer et al. (62) also acknowledge that the stiffness established during preactivation determines the elastic behaviour of the muscle-tendon complex. Additionally, as elastic energy is partly stored in the contractile elements of the series elastic components of muscle (140), any increase in activation, number of cross-bridges and therefore stiffness (61), would result in more elastic energy storage and utilisation (140).

Horita et al. (77) propose that the muscle under high eccentric loading, for example during drop jumps (149), should act as a stiff spring, to absorb the high force of impact. An important mechanism for allowing the muscle to function as a "stiff spring" is to increase muscle stiffness on and before impact. Central programming and supra-spinal control could realize this preactivation, and therefore increase muscle stiffness (12;13;60;116). Avela and Komi (12) suggest that pre-landing stiffness, enhanced by central preactivation, could increase post-landing stiffness. Because a high degree of muscle stiffness is recommended for effective utilisation of elastic energy, this increased stiffness should be optimal for energy storage and subsequent release in stretch shortening cycle actions (62).

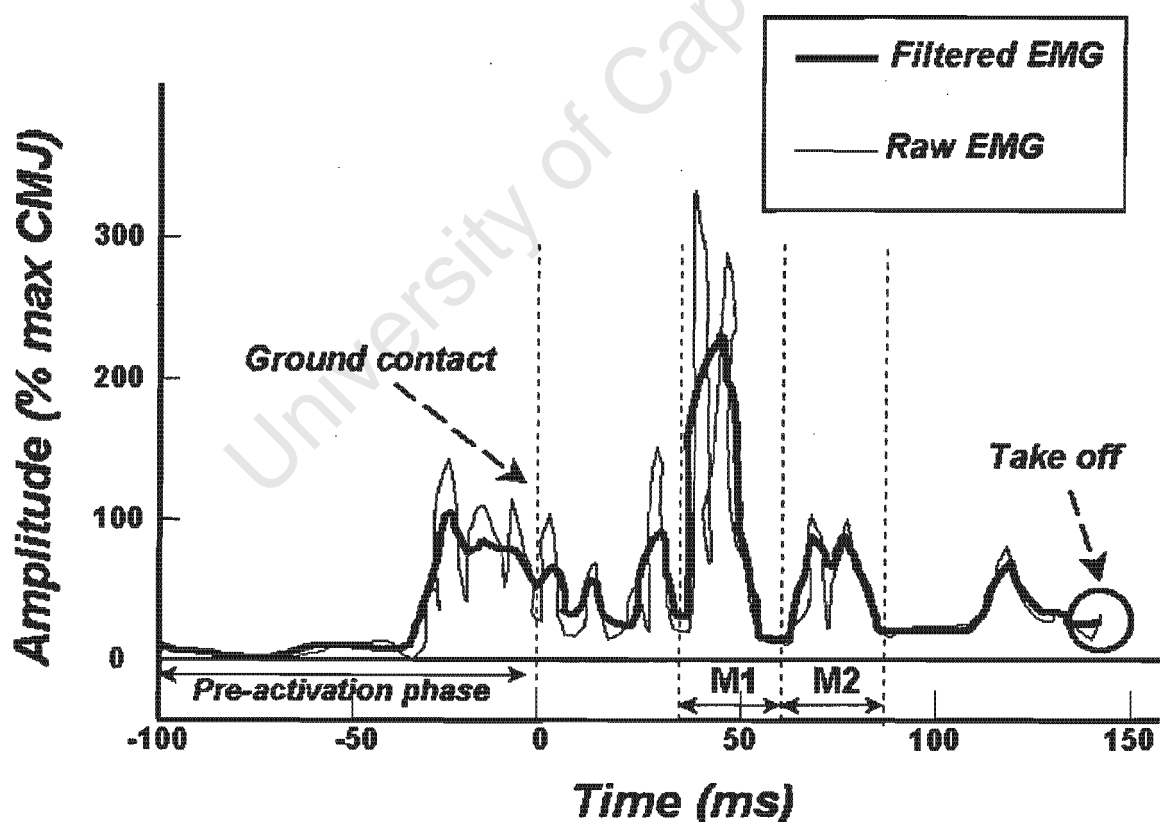
In many studies, the time-period of 100 ms before ground contact (120) or the eccentric phase (10) defines the preactivation phase in vertical jumps (7;9;11;77) (Figure 3). Dietz et al. (41) also confirm this time-period in the upper body, finding that raw EMG in the triceps brachii increased about 100 ms before ground contact during forward falls.

### ***Reflex EMG and muscle stiffness in vertical jumping***

As stretch shortening cycle activity forms an integral part of natural human locomotion, it is logical to assume that proprioceptive reflexes are also involved to some extent (93). The effect on enhanced concentric performance, as a result of prior eccentric contraction, has been attributed to a combination of both elastic energy and stretch reflex facilitation (140). It has been suggested that if one were able to associate muscle stiffness with reflex

responses, one might be able to better understand the importance of both peripheral and central mechanisms and their adaptations to training (123).

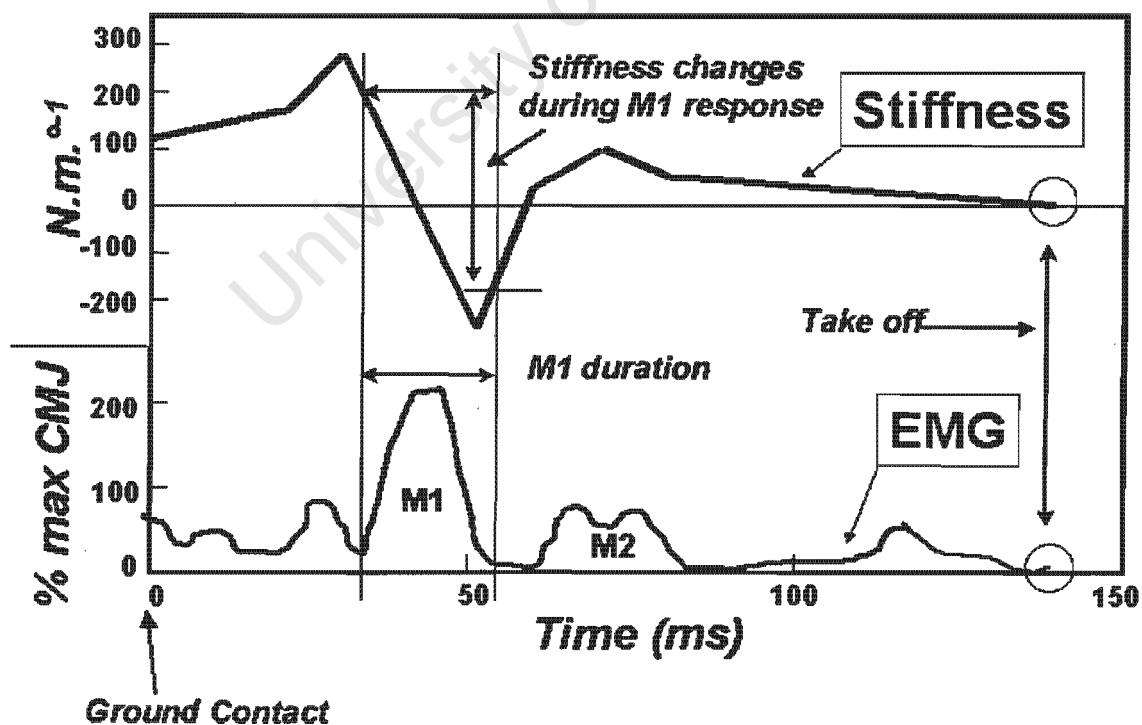
The neural control of muscle function consists of both central and reflex mechanisms (58). Gollhofer et al. (58) proposed that muscle activation during the eccentric phase of stretch shortening cycle actions includes a stretch reflex contribution, which may increase the effectiveness of the stretch shortening cycle exercise. Trimble et al. (142) suggest that spindle afferent discharge, as a result of eccentric stretching during the stretch shortening cycle, should assist and synchronise activation of motor units in the involved muscles. They additionally inferred that these afferent signals are produced primarily during the eccentric- and eccentric-concentric transition phases of the stretch shortening cycle. Faster eccentric stretching of the active muscles leads to a stronger electrical activation of these muscles, which increases stiffness (30). This increase in stiffness seems to be an outcome of reflex activity in conjunction with pre-programmed innervation (41;61).



**Figure 3:** Raw and filtered rectified EMG of the vastus lateralis muscle of one subject during a drop jump. Represented is the preactivation EMG (100 ms before ground contact). Modified from Horita et al. (77).

Koceja and Kamen (88) have shown that in reflex function both spinal and supraspinal influences are involved. Measures of reflex EMG related to neural activity in this case would be the short latency (M1) component (12;58;77) that is indicative of the involuntary monosynaptic stretch/spindle-reflex, the involuntary medium latency (M2) component (58;77) and voluntary long latency (M3) component (58;147); the latter both being supraspinal polysynaptic reflexes. All these reflexes contribute to increasing muscle stiffness and facilitating reflex contraction. The area under the EMG-time curve has previously been used to quantify these stretch reflex contributions according to their latencies (12;77).

Gollhofer et al. (58) identified at least 4 different phases of high muscle activation following submaximal forward drops in the upper body. The fourth phase could not be accurately defined, but the first three phases occurred at  $28.3 \pm 4.2$  ms,  $65.8 \pm 11.0$  ms and  $106.5 \pm 7.2$  ms after ground contact respectively, which essentially correspond to the M1, M2 and M3 components (58;147). M1 and M2 components have also been previously defined in the lower body having an approximate latency of 30 ms and 60 ms after ground contact respectively (Figures 3 and 4) (41;77).



**Figure 4:** A representation of the muscle stiffness changes during the M1 response during a drop jump. Modified from Horita et al. (77).

In the study of Horita et al. (77) a dynamic measure of instantaneous stiffness was calculated modelled on the equation:

$$\text{Instantaneous stiffness} = [M(n+1) - M(n-1)] / [A(n+1) - A(n-1)]$$

Where:

- **M** = moment,
- **A** = joint angle and
- **n** = corresponding  $n^{\text{th}}$  (number) frame.

The data showed (Figure 4, upper part) that stiffness increased before and during impact with landing, which corresponds to preactivation. After the initial increase in muscle stiffness during the eccentric phase on landing, a yielding phase occurs, where the muscle stiffness decreases because of the sudden increase in the detachment rate of actin-myosin cross-bridges (77). This is then followed by compensatory stretch reflex activation and a matching increase in muscle stiffness. Fukashiro et al. (57) advocate that when muscle fibres are actively lengthened, detachment and reattachment of the actin-myosin cross-bridges are unavoidable. Komi (93) states that this “yielding” of the activated actin-myosin cross-bridges can be prevented by intense muscle activation. Furthermore, the stretch reflex mechanism is highly sensitive to length- and force changes in the muscle-tendon complex and is therefore probably the most effective regulator in monitoring and guiding this process of preventative muscular activation (93). It may be argued that these stiffness changes i.e. preactivation and reflex compensation, exist to prevent muscular collapse on impact and also contribute towards an efficient, subsequent concentric push-off phase (77).

Asmussen and Bonde-Peterson (61), however suggest that a simple myotatic reflex on its own would be too slow to regulate and absorb the forces upon landing. In support of this rationale, Melville-Jones and Watt (113) propose that the first functionally effective reflex contraction after a sudden active stretch occurs only after ~120 ms. This third reflex (M3) component, or voluntary long-loop reflex response, generally has a task-dependent latency of  $\pm 120 - 180$  ms and represents the cognitive interaction between the cerebral cortex and mechanoreceptors to initiate voluntary movement (147). This reflex is a voluntary reflex and is developed through conscious repetition of the task and converted to unconscious programming, as the result of repetitive training (147).

In summary, the precise details of reflex regulation of stiffness during vertical jumping have not been fully explained. However, it can be concluded that the combination of stretch reflexes have sufficient time to contribute significantly towards performance enhancement during stretch shortening cycle movements by increasing the force- and power output of the working muscles (93).

### ***EMG ratios in vertical jumping***

Fukashiro et al. (56), using squat jump and countermovement jump techniques, performed an *in vivo* measurement of achilles tendon (AT) loading during jumping. They performed this measurement by implanting a buckle-type transducer under local anesthetic around the AT. The subject then performed various jumps, while force and mechanical work around the AT was measured.

The AT elasticity during jumping was measured via direct force measurement and calculation of length changes. The changes in tendon length were calculated on the assumption that tendon stiffness was constant. A constant was used based on the physiological tendon structure and integrated with the registered force to calculate the change in length (56).

Comparisons between three types of jump i.e. squat jumps, countermovement jumps and repetitive submaximal hops were made. Even though the hops were performed submaximally, the maximal AT force generated in the hops was far greater than both the squat and countermovement jump conditions. They attributed this difference to the instantaneous flexion of the knees and hips upon contact, which actively stretched the gastrocnemius and soleus muscles.

This increased eccentric stretching of the muscle increased the force generated in the AT. Another interesting finding in this study was with the measurement of EMG. The AT force (ATF): EMG ratio for each condition was measured, by dividing the ATF values (N) by the EMG values (mV), and getting a relationship ( $\text{N.mV}^{-1}$ ). Interestingly, the ATF: EMG ratio was greater in the hops than in any of the jumps (56).

According to the force-velocity relationship, force output decreases with increasing muscle shortening velocities and therefore neural drive, and correspondingly EMG activity,

increases to compensate (56). This statement is made on the assumption that the electrical activity of muscle is related to neural drive or central activation (108). In accordance with this paradigm, the force: EMG ratio should therefore decrease with high muscle shortening velocities (56). However, Fukashiro et al. (56) found the inverse of what was expected i.e. a more than twofold increase in force: EMG ratio in hopping compared to squat jumps and countermovement jumps, despite the fact that the shortening velocity was higher. They attributed this finding to tendon elastic return. An increase in this ratio therefore seemingly indicates increased tendon elastic recoil contribution, and therefore utilisation of elastic (strain) energy (137).

Svantesson et al. (139;140) found similar results. They also found greater concentric torque preceded by eccentric contraction, and showed a reduced or unchanged EMG when compared to pure concentric contraction. These findings imply that tendon elasticity was the major contributor to this stretch shortening cycle potentiation of performance (137). This conclusion arises from the occurrence of a shift in the force-velocity curve to higher forces at high speeds of contraction, or higher speeds of contraction with more force generated by the muscles, along with a reduction in EMG or muscle recruitment (137). Bosco et al. (30) showed that the EMG: force ratio was its highest during the pure concentric squat jumps, and decreased with jumps involving pre-stretch. They further rationalised that with a lower EMG: force ratio, there would greater elastic contribution, as more force was generated with relatively less muscle activation (30).

Another measure of elastic contribution could be the eccentric: concentric EMG ratio used by Kyröläinen et al. (105). With increased eccentric muscle activity, there is an increased activation of muscle and corresponding EMG (62). If the EMG during the concentric phase is reduced or remains unchanged together with equal or higher work being performed respectively, one can assume that the concentric performance has a higher elastic contribution (74;139). If the eccentric EMG remained the same or increased, and the concentric EMG was reduced or remained constant respectively, then the eccentric: concentric EMG ratio would increase. Because there is a large increase in eccentric EMG compared to concentric EMG during stretch shortening cycle movements, it has to be assumed that there is significantly more neural activation during the eccentric phase and therefore also more strained cross-bridges (9). This dominating eccentric EMG may increase muscle stiffness and thereby increase the elastic contribution of the muscle-

tendon complex to the ensuing concentric movement (105). The eccentric phase of stretch shortening cycle movements modifies not only the force output in the subsequent concentric phase, but also alters the neural (EMG) and metabolic (elastic vs. chemical energy) contributions to the potentiation in performance (90).

### ***Inherent jumping technique variations***

A methodological concern of testing performance in the vertical jump is the standardization of jumping technique used (21). The main problem encountered is in the drop jump, where the depth of the drop jump and the kinematics of the drop jump curves are variable amongst individuals.

Bobbert et al. (21;22) identified different drop jump strategies i.e. a 'counter' drop jump and 'bounce' drop jump strategy. The 'bounce' jump group showed enhanced jump potentiation compared to the 'counter' group, which did not show potentiation (22). These jump styles showed larger and lesser movement amplitudes and longer and shorter contact times respectively. The researchers found that the choice of jumping style was arbitrary and not ascribed to anthropometrical differences (21). It has subsequently been established that the type of drop jump technique strongly affects the mechanical output of the muscles (17) and should therefore, also affect the kinematics and dynamics of the stretch shortening cycle.

In accordance with this, if the moment around a joint declines more rapidly, as was found in the 'bounce' group compared to the 'counter' group, the series elastic components might shorten faster (22). Therefore, as the majority of the induced shortening is performed by the recoil of the series elastic components, the muscle fibres would be able to contract more forcefully and powerfully in the 'bounce' group, due to improved force-velocity and length-tension relations (22).

Bobbert et al. (22) found the greatest potentiation in their 'bounce' drop jump group and the lowest in their 'counter' group. Therefore it was concluded that the magnitude of pre-stretch potentiation of concentric performance improves with the speed at which it is performed, and also decreases the longer the eccentric-concentric transition (22).



Thys et al. (141) showed that the utilisation of elastic energy was greater in small compared to large amplitude jumping movements. Bosco et al. (29) also found that elastic energy restitution and therefore contribution to the positive phase accounted for ~50% in the small amplitude jumps and ~30% in large amplitude jumps.

Another possibility, which might explain this enhancement of performance of small amplitude jumps over large amplitude jumps could be related to cross-bridges. Rack and Westbury (126) suggest that when the movement amplitude becomes larger, it is possible the cross-bridges that had gained tension by being actively stretched, could have broken their links and rejoined at a lower tension thereby losing elastic energy. This reduction in tension stored by the cross-bridges would create a lesser elastic energy return than would have occurred had they remained attached as found in short range movements (126).

Rack and Westbury (126) reasoned that large amplitude movements are also met with frictional resistance and that the muscle itself has to perform a considerable amount of work to maintain the movement, whereas short range movements are essentially elastic in nature and the muscle therefore expends less energy.

Bobbert et al. (17) state that vertical jump height is further constrained by anatomical and mechanical limitations. To prevent injury, the angular velocities at the hip and knee joints need to be low at full extension, and the contribution of both hip- and knee joint to vertical acceleration and jump height is over long before maximal extension (17). This indicates that muscle coordination, or patterns of muscle contraction might be an additional factor concerning jumping ability (17).

Bobbert et al. (17) propose that it is only possible to release more elastic energy if the muscles are able to produce a higher power output, and if the distance over which the muscles contract is identical. It is logical to assume that this form of confounding might also be applicable for future testing in countermovement jumps. It may therefore also be prudent to standardize countermovement depth and range of motion.

In summary, these findings suggest that it is important to standardize the technique used to compare kinematics and other performance parameters accurately.

### ***Physiological effects on vertical jumping ability***

Another methodological consideration in the comparison of squat, countermovement and drop jumps is the subject selection. Bosco and Komi (26) showed that the ageing process affected both the elastic behaviour of muscle and reflex potentiation. The effect of pre-stretch improvement in concentric performance increased up until the age of ~20 - 30 years old and thereafter decreased almost linearly with ageing. Bosco and Komi (26) suggested that this decrease in performance could be due to a progressive reduction in fast twitch (FT) fibres associated with ageing. This age-related trend is similar in males and females, except that this decline in potentiation seems to start earlier in females, possibly due to the increase in fat mass associated with puberty (26). When selecting a population group for research, this ageing and gender effect on performance should be taken into consideration. It may also be assumed that the ageing process adapts the neuromuscular system to protect the body against high stretch loads (26).

Aura and Komi (11) compared the mechanical efficiency of males and females during stretch shortening cycle exercise. They found that during low stretch loads females had an enhanced capacity to utilise elastic energy via pre-stretch compared to males. They also found that males had an increased capacity to utilise pre-stretch with higher stretch loads. They attributed this gender difference to possibly greater muscle strength and thus muscle stiffness in males than in females (11). Differences in body dimensions might also have contributed to the discrepancy between elastic energy utilisation of males versus females (89).

Bosco and Komi (24) found a relationship between muscle fibre-type composition and various vertical jump variables, particularly vertical jump performance. FT fibres were positively related to both countermovement jump ( $r = 0.48$ ;  $p < 0.01$ ) and squat jump ( $r = 0.37$ ;  $p < 0.05$ ) performance. It was therefore suggested that a subject with a higher percentage of FT fibres would have a faster rate of force development and thus be able to move a certain load faster, than individuals with a higher proportion of slow twitch fibres (ST) (24). These results were supported by another study, where vertical jump height was positively correlated with FT fibre distribution and negatively correlated with the rate of force production (72).

Viitasalo and Komi (145) had conflicting results when measuring the force-time characteristics of individuals with different fibre type dominance. They showed that the rate of force development was positively related to the dominance of ST fibres in the vastus lateralis muscle, in contrast to the results Bosco and Komi (24).

Another study by Bosco et al. (29) concluded that ST and FT fibres have different viscoelastic attributes that allow them to benefit differently from stretch shortening cycle movements depending on the speed of the movement. The depth or amplitude of the jumps plays a major role in this process. For example, subjects with a higher proportion of ST fibres had a higher potentiation than the subjects with a higher proportion of FT fibres in the large amplitude jumps, and the FT fibre group had a higher potentiation effect than the ST group during the smaller amplitude jumps. These findings were related to the coupling time of the actin-myosin cross-bridges, which varies between FT and ST fibres. Therefore during large amplitude jumps, the ST fibres were able to retain their stored energy longer without cross-bridge detachment and subsequent loss of potential energy.

Viitasalo and Bosco (144) also performed a study regarding muscle fibre type and its influence on vertical jump performance and found that subjects with a higher proportion of ST fibres had a better utilisation of stretch shortening cycle ability when compared to subjects with a higher proportion of FT fibres. They compared drop- and countermovement jump height as a percentage of their squat jump height. The ST group had a greater percentage improvement in jump height than the FT group when utilising the stretch shortening cycle, indicating an increased stretch shortening cycle ability. However, when examining the data, the FT group had a significantly higher squat jump height than the ST group (144). This significantly affects the percentage increase in jump height as a result of the stretch shortening cycle, and confirms that fibre type distribution may contribute to stretch shortening cycle ability.

## ***Training effects on vertical jumping***

After a 12-week strength-training programme, Häkkinen and Komi (73) found an increase in both squat jump heights and countermovement jump heights. However, the potentiation of the stretch shortening cycle, represented by the equation (***Potentiation = (CMJ-SJ)/SJ x 100%***; p 15) remained relatively unchanged. The authors proposed that the pre-stretch speed during the eccentric contractions in the training programme utilised in this study might have been too slow to develop stretch shortening cycle ability. They attributed the improved jumping ability in both squat jumps and countermovement jumps to an increase in maximal force generated by the muscles. Increased jumping ability has been associated with increases in strength of the involved musculature (134). This could possibly also be explained by enhanced FT motor unit activation during the early phases of resistance strength training (73).

Häkkinen and Komi (73) additionally found improved stretch shortening cycle ability in drop jumps from drop heights of 20 - 100 cm. The differences were most pronounced at the higher heights (80 - 100 cm), which imposed a greater pre-stretch load. Strength training supposedly increases musculotendinous stiffness (115;153). These findings contradict the findings of Walshe and Wilson (149) that the more compliant subjects have greater stretch shortening cycle potentiation during high eccentric loading.

Walshe and Wilson (149) studied the relationship between musculotendinous stiffness and various stretch shortening cycle actions in the form of vertical jump variations. They normalised stretch shortening cycle performance for jumping ability (21) modelled on the equation:

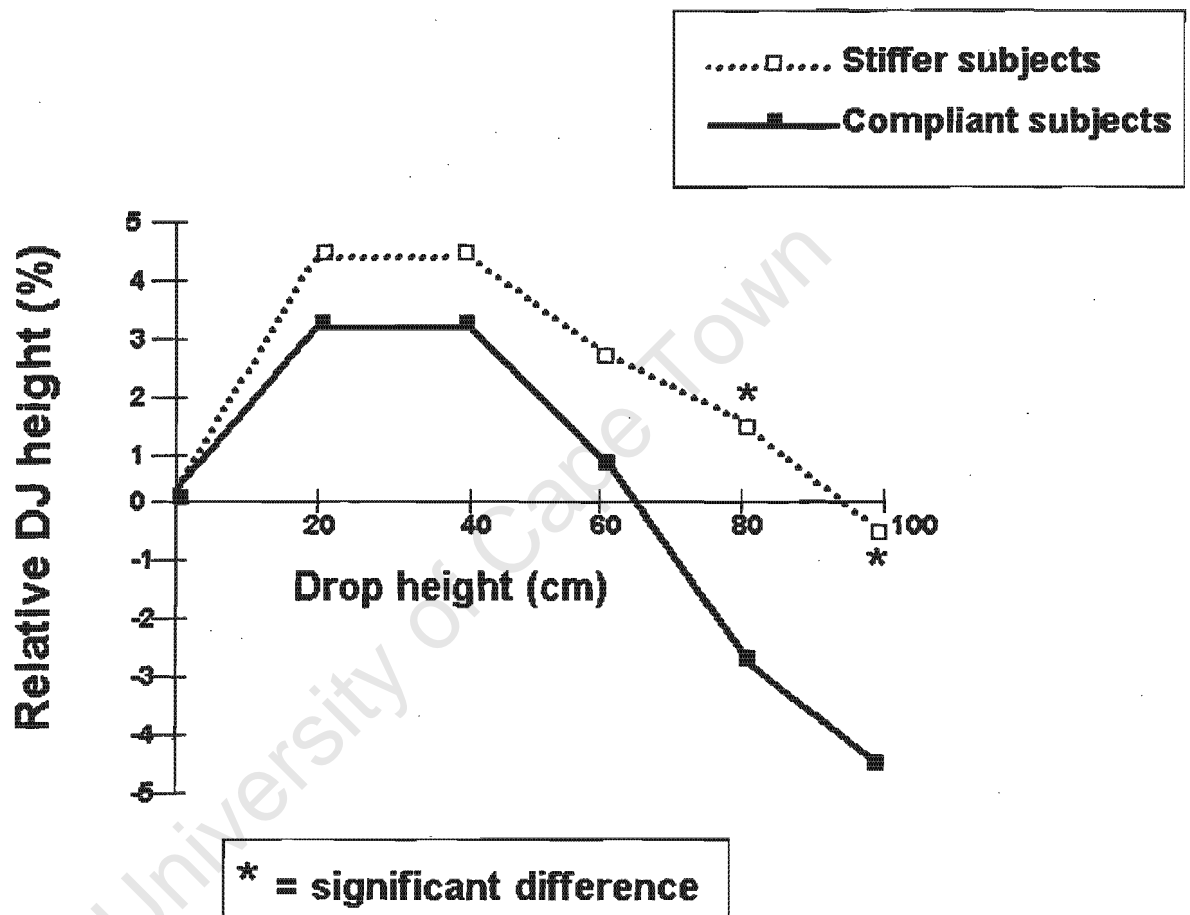
$$\text{Relative DJ} = (\text{DJ} - \text{CMJ}) / \text{CMJ} \times 100\%$$

Where:

- **DJ** = drop jump height, and
- **CMJ** = countermovement jump height

They found that the stiffer subjects did not perform as well as the more compliant subjects in high eccentric loading conditions (Figure 5). They contributed this lesser jumping ability to Golgi tendon organ reflex inhibition. According to Walshe and Wilson (149) it seems

that the more compliant subjects had a greater capacity to absorb high eccentric loads, and therefore exhibited enhanced performance capabilities, when compared to stiffer subjects utilising the stretch shortening cycle under these conditions. Watsford et al. (152) found similar results using the same drop jump protocol design compared with musculotendinous stiffness measured on the triceps surae muscle group.



**Figure 5:** Using the percentage difference between countermovement and drop jumps, a comparative analysis is shown between the nine most stiff and the nine most compliant subjects. A significant difference was found between DJ80 and DJ100 groups. Modified from Walshe and Wilson (149).

Golgi tendon activation, and therefore inhibitory drive, increases exponentially with increasing stretch load (58). The improvement in vertical jump height from high stretch loads, after strength training, may relate to reduced Golgi tendon organ inhibitory drive via neural adaptation to this form of training (73;155). It is possible that strength training caused adaptations in the neuromuscular system, which reduced inhibitory drive and improved the subject's ability to tolerate high eccentric loading, and thereby improved jump

height (73). Heavy resistance training could thereby improve high stretch-load tolerance and increase maximal force (72;73). This suggests that jumping ability could possibly improve by means of specific resistance training, as a consequence of an improved capacity of the individual to release energy, and the improved coordination of the involved muscles (17).

The effect of training is also evident in the utilisation of elastic energy when comparing athletes of different training backgrounds. This can be seen in the study by Häkkinen et al. (72), where they compared powerlifters, body builders and wrestlers. The athletes were tested during normal squat- and countermovement jumps, as well as performing the jumps with additional 20, 40, 60, 80 and 100 kg loads. The barbell was placed on the shoulders and gripped tightly by the hands during all the jumps.

With wrestlers having an extensive explosive power training background at high resistance loads, it was noticeable that they showed significantly higher differences between squat- and countermovement jumps during resistive loading, i.e. 0, 40 and 100 kg loads. The authors therefore concluded that the wrestlers had a more efficient utilisation of elastic energy during stretch shortening cycle movements. Despite having similar maximal strength values, it was concluded that their training history could have had a significant influence on the various components of neuromuscular and physiological performance, such as the rate of force production and utilisation of elastic energy between the groups (72).

It is also possible that endurance training can alter the excitability of motor neurons, and the muscle receptor response to stretch (123). Pérot et al. (123) suggest that an increased stiffness of endurance-trained muscles might be the cause of this response. They do however state that one still needs to confirm this conclusion in human muscle after endurance training. Kubo et al. (97) found an increased stiffness in endurance trained runners, which seems to support this possibility.

Plyometric training enhances eccentric muscle function by increasing the rate of force production (154). Because plyometric training enhances the rate of eccentric force development, it could therefore also have a significant effect on increasing elastic

(mechanical) energy storage in the muscle-tendon unit (154) and thereby influence stretch shortening cycle performance.

### ***Biomechanical simulation models in vertical jumping***

Simulations and biomechanical modelling techniques have also been used to study the dynamics of the human muscle-tendon complex during voluntary movement (4;14;19;20;128). This approach has been used mainly because of the difficulties in evaluating the mechanical behaviour of these structures *in vivo* (100).

One should interpret vertical jump height experiments with caution, when trying to evaluate the utilisation of elastic strain energy (4). To do this accurately, one needs to quantify individual, time-varying forces and shortening velocities on the various elastic and contractile muscle components (4). In accordance with this Anderson and Pandy (4) designed a study to provide a better understanding of how elastic tissues enhance muscle performance during vertical jumping. They used EMG, force-plates and a motion analysis system to measure limb position. Retro-reflective markers were used to measure joint dynamics and kinematics in the cinematographic analysis. In addition they used a control musculoskeletal simulation model previously developed by Pandy (in (4)). They first confirmed this model's ability to replicate the major features of each jump and then analysed the elastic contributions in the various jumping conditions. The series elastic component, tendon and parallel elastic elements, which essentially form the muscle-tendon complex, contributed ~35% of the energy transferred to the skeleton. The remaining ~65% of the energy contribution came from the contractile elements (4).

Although Anderson and Pandy (4) could not identify if the contribution of the elastic energy was the main factor determining the differences in jump height, they showed that countermovement jumps were superior in converting potential energy into elastic (strain) energy than squat jumps and concluded that this ability improved the efficiency of the jump. They further looked at the effect of tendon compliance on elastic energy storage and found that increased compliance led to an increased elastic energy contribution to the vertical jumps. However a shortcoming of their model was that it underestimated the amount of energy that the contractile tissues could deliver and contribute to both squat jump- and countermovement jump height.

Of all the muscles tested during vertical jumps, only the vastii and hamstring muscles developed larger forces in countermovement jumps compared to squat or static jumps. The calf musculature i.e. soleus and gastrocnemius did not show any large increases as they were mainly activated during the concentric propulsion phase of the jumps. The tendons of the vastii and hamstrings are relatively short and stiff, which is not conducive for a major increase in elastic energy storage during the countermovement jumps (4). However, using a different research technique, i.e. *in vivo* ultrasound measurement, Kubo et al. (101) conversely found that the tendon and tendinous structures in vastus lateralis were considerably compliant and seemingly contributed to an increased elastic energy storage and usage in countermovement jumps.

The main conclusion of the abovementioned study (4) was therefore that when jumping is preceded by a countermovement or fast eccentric stretch, the storage and utilisation of elastic strain energy leads to a more efficient, rather than significantly higher jump (4). During stretch shortening cycle movements, the increased stored elastic energy during the eccentric contraction occurs mainly due to the lengthening of the series elastic elements at the expense of lengthening of the contractile components. This storage of elastic energy during the eccentric phase simply allows the ensuing concentric contraction to be performed more efficiently, because it reduces the work performed by the contractile components (18;128).

### ***Mechanical efficiency in vertical jumping***

Shorten (132) confirms that elastic energy storage and utilisation during the stretch shortening cycle results in a more efficient performance, and that it also makes the muscles involved perform more effectively. In other words, the recoil of the elastic tissues performs part of the work of raising the body (28;36), which thereby lessens the work performed by the active muscles and reduces the metabolic cost of the movement, i.e. improves its efficiency (7;9;28;33;98;128).

This refers to the concept of mechanical efficiency. Mechanical efficiency is commonly used to describe the efficiency of man in motion (10) or the economy of movement, i.e. the amount of mechanical or external work performed as a percentage of its energy cost (11;90;104). The main factor contributing to improved muscular efficiency during stretch



shortening cycle movements is the elastic properties of the muscle-tendon complex (14). Mechanical efficiency has been shown to be significantly greater following the stretch shortening cycle muscle action (104). Roberts (128) proposes that due to the elastic nature of tendons, combined with different shortening velocities of tendon and muscle respectively, the muscles might operate at lower shortening velocities, where the force is generated at a lower metabolic cost.

Bosco et al. (23) found that mechanical efficiency increased with increasing velocity of jumping and therefore reasoned that energy may be stored in the series elastic components during pre-stretch for subsequent use in the concentric contraction. Kyröläinen and Komi (104) also found an increased mechanical efficiency with increasing eccentric stretch velocity, which partly supports this finding. Aura and Komi (9;11) found a positive correlation between pre-stretch intensity, elastic contribution and mechanical efficiency. Both elastic contribution, and mechanical efficiency increased with increased pre-stretch intensity.

Belli and Bosco (14) found that when utilising stretch shortening cycle movements versus pure concentric movements in the triceps surae muscle group, the contractile component efficiency was markedly improved during the stretch shortening cycle movements. They found a significant increase in mechanical efficiency in rebound lifts ( $30 \pm 5\%$ ) compared to pure concentric lifts ( $18 \pm 3\%$ ) in the triceps surae muscle group. Work attributed to the series elastic components during rebound lifts contributed  $\sim 40\%$  of the external work of the center of gravity. When compared to the pure concentric contraction, where the series elastic component work was negligible, it accentuates the value of elastic energy restitution in stretch shortening cycle movements. They further stated that the cross-bridge stiffness during stretch shortening cycle movements such as those used in this study, was higher than tendon stiffness, and therefore the major part of the elastic energy was stored in, and released from, the tendon (14).

Aura and Komi (10) have shown that the mechanical efficiency of isolated eccentric exercise is much greater than concentric exercise and that this improved even more with higher mechanical work. They also showed that the efficiency of isolated concentric exercise showed the opposite trend i.e. a decrease in mechanical efficiency with increased mechanical work.

Bosco et al. (28) found that non-rebound, or squat jumps, had a mechanical efficiency of 18% and rebound jumps, or countermovement jumps, had a mechanical efficiency of 28%, which was significantly higher than non-rebound jumps. As the positive work performed by the muscle-tendon complex was significantly greater than the potential energy that could be stored during the eccentric contraction, Bosco et al. (28) suggested that there were additional factors in addition to the elastic recoil contribution that realized this increase in mechanical efficiency.

Komi (91) states that improvements in force, speed and power after stretch shortening cycle activity also imply that mechanical efficiency is enhanced. The force: EMG ratio, as used in the study of Fukashiro et al. (56), could also possibly be used as a measure to confirm improved economy of positive work. The increase in force, with a reduced EMG or muscle activation, would therefore imply an improvement in mechanical efficiency (23). Bosco et al. (23) found a reduction in the EMG: force ratio in rebound jumps utilising the stretch shortening cycle, compared to pure concentric squat jumps, and also small amplitude rebound jumps, compared to large amplitude rebound jumps. They proposed that to develop the same amount of force in rebound jumps less neural activation (EMG) is required, and therefore a certain amount of work is derived from elastic origins. These measures seem to be related as the EMG: force ratio followed a similar pattern to mechanical efficiency in the abovementioned study.

## **BENCH THROWS**

Newton et al. (119) used a bench throw movement to study stretch shortening cycle activity, as it was a 'natural' multi-joint movement frequently used in resistance training, and it was analogous to the vertical jumps used in lower body stretch shortening cycle studies.

The upper body version of the vertical jump testing procedure is the pure concentric bench throws and rebound concentric bench throws. During this technique one measures vertical ground reaction force (119;154). For standardization of grip, subjects were required to maintain the shoulders in a 90° abducted position to ensure consistency of shoulder and elbow angles, and the performance of the movement (119). For the pure concentric throws, the bar was rested  $\pm 1$  cm above the chest on the bottom stoppers.

From here subjects were instructed to push and throw the bar up vertically trying to obtain maximum height. For the rebound throws they were instructed, starting from an extended arm position, to quickly lower the bar to the chest and immediately reverse direction and throw the bar up vertically in an attempt to obtain maximum height. EMG was measured throughout using three muscle groups i.e. anterior deltoid, the sternal portion of pectoralis major and the long head of triceps brachii (119).

Newton et al. (119) found similar force-velocity-power kinematics to that found in isolated muscles, single-joint movements, and vertical jump performance in the lower body. This indicates that the performance of upper body explosive movements is very similar to that of the lower body. Newton et al. (119) found that the stretch shortening cycle did not, however, enhance peak velocity or height thrown (Table 2), but improved the early phase of the concentric movement, after which this augmentation in muscle performance diminished. They ascribed this early enhancement to the recovery of elastic energy and stretch reflex potentiation. The augmentation of the concentric performance parameters i.e. power, force and velocity occurred primarily within the initial 50 - 80% of the concentric movement (119).

**Table 2:** Evidence of stretch shortening cycle potentiation of muscle performance in the upper body musculature.

Stretch shortening cycle potentiation of performance in the upper body				
Reference	Variable	Test	Measured	Potentiation
Newton et al. (119) Male Exercise Science students (n = 17)	Throw height			
	15% 1RM	PCBT RBT	64 ± 26 cm 74 ± 11 cm	-
	30% 1RM	PCBT RBT	42 ± 16 cm 36 ± 5 cm	
	45% 1RM	PCBT RBT	25 ± 6 cm 23 ± 5 cm	
	60% 1RM	PCBT RBT	17 ± 5 cm 13 ± 3 cm	
	75% 1RM	PCBT RBT	9 ± 3 cm 8 ± 2 cm	
	90% 1RM	PCBT RBT	5 ± 3 cm 4 ± 3 cm	
	Mean	PCBT RBT	27 ± 24 cm 23 ± 25 cm	
Wilson et al. (156) Male experienced weightlifters (n = 12)	Weight lifted	PCBP RBP	125 ± 25 kg 143 ± 28 kg	-
Wilson et al. (156) Male experienced weightlifters (n = 12)	Impulse	PCBP and RBP	-	16 ± 4 %
Wilson et al. (153) Male powerlifters (n = 9) Control group	Weight lifted	PCBP RBP	116 ± 13 kg 129 ± 14 kg	-
Wilson et al. (153) Male powerlifters (n = 9) Experimental group	Weight lifted	PCBP RBP	118 ± 24 kg 133 ± 25 kg	-
Doan et al. (42) Moderately trained males (n = 9)	Weight lifted	RBP SME	97 kg 101 kg	-

**Index:** PCBT = pure concentric bench throws, RBT = rebound bench throws, PCBP = pure concentric bench press lifts, RBP = rebound bench press lifts, SME = supramaximal eccentric loading

Other upper body techniques have also been used. Wilson et al. (153;156) used a similar testing procedure to Newton et al. (119). However, they elected to use a rebound bench press and a pure concentric bench press, where the bar was never thrown or released from the hands. They compared the potentiation of impulse and maximum loads lifted during a stretch-shortening cycle movement in the upper body to those of a pure concentric movement.

Wilson et al. (156) showed a significant improvement in the use of the stretch shortening cycle in an upper body strength lift such as the bench press (Table 2). They additionally used impulse measures, calculated over the first 370 ms of the concentric phase of the lifts, to quantify potentiation of stretch shortening cycle performance over pure concentric performance, using loads equivalent to 95% of their 1RM at each respective condition.

Augmentation of concentric performance was gauged as:

$$\% = (\text{Rebound impulse} - \text{Concentric impulse}) / \text{Rebound impulse} \times 100\%$$

Wilson et al. (153) also confirmed a functional improvement (increase in force and loads lifted) and reinforced the value of the stretch shortening cycle in upper body movements.

During resistance training a limiting factor to lifting weights is the 'sticking point', or the weakest point in a muscles range of motion due to mechanical disadvantage. It is a plausible theory that during the initial phase of concentric contraction, as part of the stretch shortening cycle during a rebound bench lift, the potentiation in average- and peak power, raises the bar beyond the 'sticking point', increasing the maximum loads lifted, compared to the maximum load lifted during pure concentric presses (119).

Doan et al. (42) studied the effects of the stretch shortening cycle in a slightly different way. They utilised the bench press movement to illustrate the effect of increasing eccentric overload on maximal concentric performance i.e. weights lifted. They tested for the standard maximal 1RM bench press, followed by a supramaximal eccentric overload. They increased the eccentric overload by utilising a weight-release device, which additionally loaded the bar during the eccentric or lengthening phase and released the additional weight at the bottom of the lift. Subjects on average increased their 1RM significantly utilising the eccentric-overload protocol compared to the standard 1RM protocol (Table 2).

#### **SLEDGE APPARATUS**

Gollhofer et al. (58;59) used a sledge apparatus to measure stretch shortening cycle function in the upper body. Their subjects lay prone and inverted on a 15°-incline sledge apparatus so that they could perform both extension and flexion in the arms. They were also strapped onto the sledge apparatus so that movement of the body was restricted. Two separated force plates, one for each arm formed part of the base of the sledge apparatus.

For the concentric test they used a 'squat'-type movement where subjects were requested to maximally push and propel themselves upward from a 90° arm-angle bottom position. For stretch shortening cycle comparison they used what was equivalent to the drop jump condition in the legs. They measured stretch shortening cycle performance utilising drop heights of 50 cm and 10 cm above their individual respective extended arm position. For drops from 50 cm subjects were instructed upon landing to push themselves back upwards as high as possible, with as short as possible transition time between eccentric and concentric phases. For the drops from 10 cm, subjects were instructed to perform the same, but to repetitively achieve a height of 70% of that reached during the drops from a height of 50 cm.

### ISOKINETIC TESTING

Isokinetic testing has also been used to test for stretch shortening cycle enhancement of performance. Svantesson et al. (137-140) used an isokinetic dynamometer to measure stretch-shortening cycle potentiation of concentric performance. They used the triceps surae muscle group to illustrate their findings. The subjects were set up on the isokinetic device in the prone-lying position with their knees fully extended.

They first tested maximal concentric torque at 120°.s<sup>-1</sup> and 240°.s<sup>-1</sup> testing velocities using a range of movement from 80° - 120° of plantar flexion (10° of dorsiflexion to 30° of plantar flexion). After completing the set repetitions and a subsequent submaximal warm-up, subjects were further tested for stretch shortening cycle potentiation. Three sets of maximal eccentric contractions from 120° - 80° of plantar flexion followed immediately by maximal concentric contractions from 80° - 120° were performed. The concentric torque was measured in both conditions within the ranges of 90 - 99° of plantar flexion and used for further analysis (138-140).

The authors consistently found significant increases in concentric torque preceded by eccentric contraction over pure concentric contractions (137-140). These increases were also more prominent at the higher speeds of movement i.e. 240°.s<sup>-1</sup> vs. 120°.s<sup>-1</sup>. Concentric EMG values were also constantly lower or unchanged during the stretch shortening cycle. This once again reinforces the notion of utilising the force: EMG or

torque: EMG ratios as indicators of increased elastic contribution to performance and improved mechanical efficiency.

## **SUMMARY**

Measurement of stretch shortening cycle muscle function has predominantly been performed on the lower limb musculature in the form of vertical jumps. In research on the upper body, the bench press action has been the equivalent of the “vertical jumps”. These methods of measurement seem to be the most functional, applicable and comparable movements compared to the activities of daily living and sports performance. Stretch shortening cycle movements show significant improvements in functional performance i.e. jump height or weight lifted, and also show significant improvements in the kinetics, kinematics and neuromuscular components involved i.e. force, power, velocity and EMG, when compared to isolated concentric actions. It would therefore be advisable to measure these components when comparing stretch shortening cycle to isolated muscle function. It would also be advisable to control for any confounding variables, such as gender and age, and to standardise testing procedures, such as range of movement, starting position, etc. for optimal comparison of data.

## **STRETCH SHORTENING CYCLE AND POTENTIATION**

### **INTRODUCTION**

In stretch shortening cycle movements, the preceding eccentric contraction positively influences the concentric contraction, as mentioned earlier in this review. One, or a combination of mechanisms can contribute to this augmentation in concentric performance associated with stretch shortening cycle movements (79). The proposed mechanisms are:

- (i) The release of additional elastic strain (mechanical) energy stored in the series elastic components of the muscle-tendon complex i.e. elastic recoil,
- (ii) The “*Pre-stretch*” condition
- (iii) The “*Force-time*” effect

- (iv) Myoelectric potentiation of the contractile elements, and
- (v) The relationship and interaction between the tendinous structures (aponeurosis and tendon) and muscle fibres i.e. interaction effects

## **PROPOSED MECHANISMS**

### **ELASTIC RECOIL**

#### ***Experimental evidence***

The first mechanism describes the elastic recoil of the tendon and tendinous structures and subsequent release of mechanical or elastic energy, based on the force and energy dynamics discussed earlier on. When the active eccentric stretch is followed by concentric contraction, the muscle-tendon complex essentially functions like a spring (107;126). Elastic energy is stored as strain energy in the muscles and tendons, and this energy is returned via elastic recoil of these tissues (2;23;34). During stretch shortening cycle activities, the muscles contract almost isometrically (127) allowing most of the elastic energy to be taken up in the tendons, as they are stretched during the eccentric phase (83). Due to the substantial decrease in force in the transition from eccentric to concentric movements, and at the expense of muscle fibre shortening, the shortening velocity of the tendons and tendinous structures increases appreciably by means of elastic recoil (48). Therefore, during the initial phase of concentric contraction almost all the shortening is induced by the elastic recoil of these structures (48).

Shadwick (131) states that elastic recoil occurs primarily through the tendons, and it is through this process that the stored elastic energy is converted back to kinetic and potential energy. The properties of tendons allow them to function as biological springs. For example, they have a relatively high modulus of elasticity, can sustain high forces, can elongate ~5% of their length (62;131), and are very resilient. These properties result in tendons being able to reclaim much of the strain energy stored during the eccentric pre-stretch when the force upon them is reduced (1;124;131).

Bobbert et al. (20) showed that tendon elastic recoil plays an important role in the work performed and power output generated during the concentric phase of one-legged jumps.



They found that the peak power output and work performed by the triceps surae muscle complex exceeded that generated by the muscles themselves. As the gastrocnemius is a dual-joint muscle, they attributed this potentiation of power and work to two mechanisms i.e. elastic recoil of the tendinous tissues and assistance from the knee joint.

Ettema and Huijing (46) state that the series elasticity of the muscle-tendon complex cannot exclusively be attributed to the tendon and tendinous structures, as muscle stiffness generated by the actin-myosin cross-bridges also makes a significant contribution (132). However, the distance that bound actin-myosin cross-bridges can be extended before they forcibly detach, limits the elastic contribution of the muscle from this mechanism (128). Alexander and Bennet-Clark (2) inferred that the capacity of the muscles themselves for storing elastic strain energy was generally too small to be useful, and that the structural composition of tendons made them better suited for storing elastic strain energy.

Many alternative theories have been offered regarding the mechanism of performance enhancement, yet most research evidence confirms elasticity as a major contributor to force potentiation during stretch shortening cycle movements (93). Storage and release of elastic energy is not however the only mechanism whereby the concentric performance of previously stretched muscle can be enhanced (33). This potentiation occurs not only because of the recoil of elastic structures, but also due to the shortening of the contractile apparatus of muscle (34). The spring action or elastic recoil of the series elastic components of muscle, only contributes to the initial phase of the concentric movement, whereafter the contractile components perform the majority of the work (134;141)

The cytoskeletal proteins nebulin and titin also have the potential to contribute to elastic recoil during stretch shortening cycle activities (107). However, their contribution has not been studied (95).

### ***Summary***

The recoil action of the tendon, tendinous structures and actin-myosin cross-bridges contributes substantially to stretch shortening cycle potentiation of performance. During eccentric contraction, these series elastic elements are stretched and store elastic energy. When the force upon them is reduced during the concentric contraction phase, they recoil elastically, release the stored elastic energy, and assist in accelerating the limb or body

part. The effect of the recoil action of the series elastic elements seems to be most prominent in the initial phase of concentric contraction. Thereafter, the contractile components prevail.

## THE "PRE-STRETCH" CONDITION

### *Experimental evidence*

Walshe et al. (150) have shown that if the concentric movement of muscle is initiated from a higher level of force, the mechanical or external work performed by the involved muscles is enhanced during the first 300 ms of shortening. They compared force outputs during concentric squats with either pre-isometric or eccentrically pre-stretched squats (150). It was shown that in both pre-isometric and pre-stretch squats, the concentric contraction was initiated at a significantly higher force level than the pure concentric squat. No significant differences were found between the pre-isometric and pre-stretch squat conditions in their initial force outputs. Both 'pre'-conditions demonstrated significantly increased concentric performance compared to the pure concentric squat. The early effects of active pre-stretch were however greatly diminished later into the concentric contraction (50;150).

Ettema et al. (47) confirm a potentiation effect on isolated muscle for all conditions when utilising pre-stretch, or the stretch shortening cycle, without any additional neurological or elastic contribution. By inducing a quick shortening phase following pre-stretch, all surplus force generated as a result of the pre-stretch was eliminated, and the influence of the series elastic elements on subsequent concentric performance was thereby reduced. Even though they had removed the effect of elastic recoil and reflex contribution, there was still evidence that the muscle's contractile components were in some way potentiated for at least 500 ms into the ensuing concentric contraction (47).

### *Summary*

When a concentric contraction is initiated from a higher force level, irrespective of the preceding contraction or muscle action, muscle performance is potentiated. It is possible that the increased muscle activation from the preceding contraction might contribute to this

phenomenon. The contribution of this mechanism is limited to the initial part of the concentric phase of stretch shortening cycle muscle actions.

## THE "FORCE-TIME" EFFECT

### *Experimental evidence*

Bobbert et al. (18) initially offered several possible influences that might contribute to countermovement jumps (stretch shortening cycle) achieving greater jump heights than squat (concentric) jumps:

- Subjects were not used to the technique of squat jumps
- There was insufficient time to develop as high a force output at the start of concentric contraction (e.g. during squat jumps), compared to eccentrically pre-stretched muscle (e.g. during countermovement jumps)
- The storage and release of elastic energy in the series elastic elements during countermovement jumps
- Increased muscle stimulation, as a result of spinal reflexes induced via the eccentric pre-stretch
- Active pre-stretching of muscle alters muscle fibre contractile properties

They compared countermovement jumps to squat jumps from various positions within the natural range of motion. The magnitude of the force at the start of the upward movement was significantly larger for countermovement jumps, than for any of the squat jump conditions (18). The greater jump height during countermovement jumps over squat jumps could be explained by the greater joint moments generated at the start of the upward movement in countermovement jumps (18). The moments generated around the hip and knee joints in the squat jump conditions were significantly lower than those produced during countermovement jumps during the early part of the concentric or push-off phase (18). This could also explain the significant increase in work shown in many other studies (30;78;150;156) during the initial part of the concentric contraction during stretch shortening cycle activities.

It appears that the potentiation during a stretch shortening cycle muscle action, in this example countermovement jumps vs. squat- or static jumps, could be explained by the

propulsion muscles of the lower limb or the leg extensors, utilising the stretch shortening cycle and developing more force prior to the onset of concentric contraction (18). In contrast, during the squat jumps, movement was initiated as soon the force generated exceeded the force required to maintain static equilibrium (18), resulting in less acceleration of body mass and therefore lower jump height.

### ***Summary***

An explanation for the potentiation during a stretch shortening cycle muscle action is that there is time to pre-tension the muscle-tendon complex and generate high levels of force, before the onset of concentric contraction. In isolated concentric contraction, movement occurs, as soon as the force levels surpass that required for maintaining static equilibrium.

## **MYOELECTRIC POTENTIATION**

### ***Experimental evidence***

Pre-stretching active muscle increases the force generating capacity of muscle at most contraction velocities (47). Stretching activated muscle might assist the effectiveness of stretch shortening cycle exercise by inducing reflex potentiation (27;77).

Gollhofer et al. (61) state that for an effective stretch shortening cycle, pre-programmed muscle activity is enhanced by stretch reflex facilitation. The stretch reflex is controlled by both muscle spindles and Golgi tendon organs, which have either a facilitatory or an inhibitory influence on subsequent concentric performance (91;140).

Dietz and Noth (40) demonstrated that stretching activated muscle evoked a noticeably larger increase in EMG activity than stretching a passive or relaxed muscle. This may be a consequence of the formation of additional actin-myosin cross-bridge connections after a quick active stretch during the stretch shortening cycle (33), thereby facilitating a stronger concentric contraction. The prior eccentric contraction could also facilitate concentric force production by mechanically activating the involved motor units (30;33). However, Walshe et al. (150) showed no significant difference in muscle activation between pure concentric contraction, and concentric muscle contraction following pre-isometric or pre-stretch conditions. On the basis of these findings they suggest that there

was minimal contribution from myoelectric potentiation to the increased work output during vertical jumps.

Komi however specifies that effective contribution of the reflex loops in the stretch shortening cycle become increasingly important where eccentric loading is high (95), and muscle stiffness needs to be regulated to meet and absorb the impact (9;12;77;121). Another important fact to consider during stretch shortening cycle activities is the rate of eccentric stretching. Stretch reflex activity increases almost linearly with increases in the rate of stretch (58;61). The rate of pre-stretch must therefore be high, as it determines the degree of stimulation of the muscle spindles and the degree of stretch reflex contribution (54). Ishikawa and Komi (83) studied subjects performing unilateral drop jumps on a sledge apparatus, and showed that eccentric EMG increased accordingly as the drop height was raised, which supports the abovementioned rationale.

Cavagna (33) states that the greater the stretching force on the elastic structures, or the greater the motor unit activation, the greater the elastic energy stored and subsequently available for use in the ensuing concentric contraction. On the basis of the abovementioned arguments it would seem as if the eccentric or active stretching phase within the stretch shortening cycle, elicits the reflex action of the neuromuscular system, and thereby effectively contributes to efficient elastic energy storage and utilisation in the muscle-tendon complex.

### ***Summary***

Myoelectric potentiation plays an important role in the stretch shortening cycle. The main effect of stretch reflex activation is to regulate muscle stiffness and preserve tension on the tendon, thereby contributing to the efficient storage and utilisation of elastic energy in the muscle-tendon complex. Pre-programmed muscle activation is also enhanced by reflex facilitation and the extent of reflex contribution is rate dependent; the faster the eccentric stretch, the greater the reflex activity. Myoelectric potentiation might also contribute to potentiate concentric force production through pre-stretch, by the mechanical or stretch activation of additional motor units or the formation of additional actin-myosin cross-bridges within already activated motor units.

### ***Experimental evidence***

The tendons and muscles function mainly as a complex unit of elastic structures to distribute forces and prevent injury, yet also operate to supply optimal mechanical functioning to the body.

Kawakami et al. (85) showed that a task-specific interaction exists between muscle fibres and their tendons and tendinous tissues. This interaction between muscle and the series elastic elements is essential in the performance of human physical activity (118). Muscle fibre length and shortening velocity depends on the functional interaction of the series elastic elements and also the dynamics imposed upon the muscle-tendon complex (48).

During muscle contraction tendons are stretched by a specific amount depending on their compliance (67). Even though it has been known that tendons are relatively compliant, the significance this has on muscle fibre and spindle dynamics has not received much attention (67). A reason for the poor understanding is that the mechanism is complex and difficult to measure *in vivo*.

Besides the release of elastic energy and other potentiation effects associated with stretch shortening cycle movements, muscle fibre dynamics also contribute to the stretch-induced work enhancement (48). Tendon compliance has been implicated as an important factor in determining the shortening velocity of muscle fibres during stretch shortening cycle activities (98). Roberts (128) claims that tendons can significantly augment muscle performance during human movement, because the muscle-tendon complex lengthens and shortens at velocities that would otherwise be detrimental to the working of the muscle fibres were they functioning on their own.

#### **(a) The force-velocity and length-tension relationship of muscle**

For a given muscle-tendon complex, when the tendinous structures are lengthened following the application of sufficient force, as would occur during an eccentric stretch, the muscle fibres are relatively shorter by comparison (79;128). This occurs because the

muscle fibres contract to resist the stretch. For example, muscle fascicle length is significantly shorter during maximal isometric contraction of the tibialis anterior muscle ( $53.8 \pm 2.9$  mm) than at rest ( $71.9 \pm 1.9$  mm) (127). In support of the abovementioned reasoning it was further found that during eccentric contractions, muscle fascicle length was not significantly different from that found during isometric contraction (127).

Ishikawa and Komi (83) analysed the behaviour of the tendons and fascicles of the vastus lateralis muscle during drop jumps from varying heights conducted on a sledge apparatus. They showed that the interaction between tendons and fascicles was also affected by pre-stretch loading. For example, with increasing drop height, which increases the eccentric loading on the muscle-tendon complex, the muscle fascicles became stiffer and stretched less, whereas the tendinous tissue stretched more (83). In other words, when the active muscle fibres are rapidly stretched at high speed, they remain at constant length or stretch very slowly, hence transferring the majority of the stretch to the tendinous structures (67;85;128).

In the resulting high-speed concentric contraction, the converse is also true, i.e. the relative length of the tendinous structures decreases substantially more than the relative length of the muscle fibres (79). This occurs as a result of the elastic recoil of these tissues together with the decreasing joint angle. Tendon length rapidly shortens during the concentric phase because of the rapid decline in exerted force (19;79). The concentric shortening velocity of the tendons surpasses that of their muscle fibres (19;20;48). This complex relationship might explain why the force capabilities for a given velocity improves, due to a relatively increased fibre length and lower velocity of contraction of these muscle fibres (19;22;79;124;150;151).

Kubo et al. (100) also reproduced this phenomenon during stretch shortening cycle muscle function using the gastrocnemius muscle of the lower limb. They found that during the initial phase of concentric contraction the shortening velocity of the tendon structures suddenly increased at the transition from dorsiflexion to plantar flexion. A comparatively slower muscle shortening velocity accompanied this increased shortening velocity of the tendinous structures. Reducing the contractile velocity to near isometric movement velocities could enable the muscles to generate more force due to more favourable force-velocity relations (100). Cavagna et al. (36) and Cavagna and Citterio (34) verify this

potentiation effect of the contractile components. The force developed during the concentric contraction of stretch shortening cycle movements is higher than that developed during isometric contraction for a considerable time-period following eccentric pre-stretch, and this is mainly noticeable at the greater muscle lengths (36).

Kawakami et al. (85) found that during pure concentric plantar flexion, contraction was initiated from full dorsiflexion at a greater than optimal fascicle- and therefore muscle fibre length (64 vs. 50 mm). This suggested that in the pure concentric test, sarcomere contraction started from a “disadvantaged” position, i.e. due to being slightly over-stretched they were on the descending part of the length-tension curve (85). On the other hand, with countermovement, fascicle length was measured after eccentric stretch at about 55 mm at the transition point from dorsiflexion to plantar flexion, which was much closer to the optimal length ( $\pm 50$  mm) for greater force production (85). It was therefore hypothesised that due to the more favourable position on the length-tension curve, the muscle fibres could contract more forcefully (85). It is therefore feasible that during stretch shortening cycle movements, muscle fibres are displaced less and therefore function at a more optimal length for force production (42).

Kurokawa et al. (103) compared the movement dynamics of the individual “*in-series*” components of the gastrocnemius muscle-tendon complex during countermovement jumps using ultrasonographic techniques. They divided the countermovement jump into three phases, namely a downward movement phase, an initial upward movement phase and a final upward movement phase, which terminated at push-off. The individual component’s movement dynamics differed significantly. During the downward phase, the muscle-tendon complex as a whole shortened by  $\pm 1.6\%$  (7 mm), with the muscle fascicles shortening by  $\pm 22.5\%$  (15 mm) and the tendons lengthening by  $\pm 2.2\%$  (8 mm). During the initial upward movement phase, the muscle-tendon complex length remained approximately constant, with the fascicles shortening even further by  $\pm 19.2\%$  (13 mm) and the tendons stretching by a further  $\pm 4.4\%$  (16 mm). During the final upward movement phase, which led to push-off, the muscle-tendon complex shortened rapidly by  $\pm 5.3\%$  (22 mm) of its initial length, with the fascicles shortening by  $\pm 5.2\%$  (4 mm) and tendons shortening by  $\pm 5.2\%$  (19 mm) of their initial lengths (103).



The shortening velocity of the combined muscle-tendon complex during the first two phases was minimal because of the interaction of the tendons and fascicles, and increased considerably during the final upward movement phase of push-off. During the final phase of push-off muscle shortening velocity decreased to almost zero, whereas the shortening velocity of the tendon increased substantially by recoil action and predominantly contributed to the shortening velocity incurred by the muscle-tendon complex as a whole (103). During the phase where large force production is required to propel the body upwards into the air, and the muscle-tendon complex is rapidly shortening, the shortening velocity of the muscle fibres or fascicles was shown to be very low (103). Due to an altered relationship between force and velocity, the muscle fascicles were able to generate more force (19;42;103;151).

This interaction between length changes and shortening velocity differences of muscle and tendon seems to play an instrumental role in the ability of the stretch shortening cycle of muscle function to increase joint power (103). A functional role of tendon therefore seems to be as a power amplifier during stretch shortening cycle movements, by decreasing the time over which the movement is performed (1), and by altering the force-velocity and length-tension dynamics of muscle.

#### **(b) The influence of the tendon aponeuroses**

Most skeletal muscle fibres attach to tendon plates or aponeuroses, which collect together at the respective ends of the muscle and form the tendons, which connect muscle to bone (84). The change in length of the muscle-tendon unit depends on two main factors, namely the force exerted upon it and secondly the muscle length at which the force is generated (45). The aponeurosis might contribute significantly to the functioning of the series elastic components (45;124). Kubo et al. (102) suggest that the *in vivo* elastic properties of the tendons and tendinous structures are greatly compliant and that the aponeurosis may in fact be the main contributor to this phenomenon. Ettema and Huijting (45) measured the properties of the various tendinous structures and series elastic elements of muscle. The tendon aponeurosis seems to be influenced by the force applied to the muscle-tendon complex as well as the length of muscle fibres, where tendon is not influenced by muscle length, but rather by the applied force (45).

Ettema and Huijing (45) proposed that during a stretch shortening cycle movement there might be extra energy uptake by the aponeurosis, which remains undetected. During the concentric phase of a stretch shortening cycle movement, the additional elastic recoil contribution of the aponeurosis might further add to the changes in shortening velocity- and muscle fibre length witnessed in the literature (45).

Maganaris (109), however, suggests that even though the aponeuroses are connected in series, they are affected differently to the tendon. The aponeurosis and tendon have different effective forces pulling against them. The tendons being further away from the centre of the muscle belly have the full force generated by the muscle applied to them. As one moves along the aponeurosis towards the centre of the muscle belly, less muscle fibres are exerting force upon it (109). This complicates the possible effects of the aponeurosis on muscle fibre dynamics even further. The muscle fibres that attach to the more extendable regions of the aponeurosis i.e. closer to the tendon would shorten more and the fibres that attach to the less extendable regions i.e. nearer the centre of the muscle belly would shorten less (109). This implies different force-velocity and length-tension relationships in different fibres within a single muscle, depending on their location. In contrast to Maganaris (109), Muramatsu et al. (118) however have shown that there is no difference in strain between the distal and proximal aponeurosis of the medial gastrocnemius muscle. They postulated that this finding could be explained by the systematic decrease in thickness of the aponeurosis as it extends into the belly of the muscle (118), which coincides with the gradual reduction in force applied to it as it moves inwards (109).

Differences in muscle fibre lengths between pre-stretch and pre-isometric contractions alone, cannot completely explain the enhanced concentric contraction after stretch shortening cycle movements (48). This suggests a potentiation of the contractile components of muscle (48).

Kawakami et al. (85) concluded that it was the combination of both the greater force-generating capabilities of the muscle fibres and the elastic recoil of the tendon and aponeurosis that were the main contributors to the enhanced concentric performance during stretch shortening cycle movements. They do however mention that one cannot necessarily exclude the possible effects of neural stretch-activation.

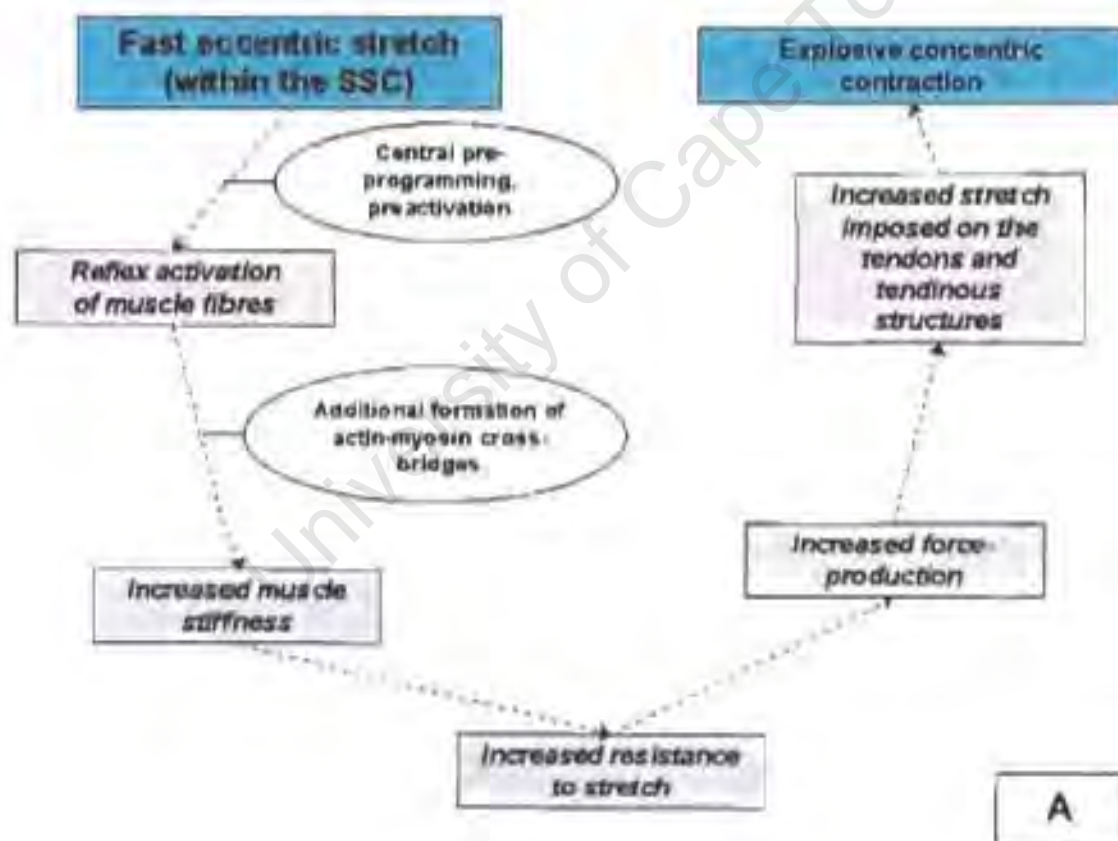
## ***Summary***

Tendon compliance plays an active role in the energy dynamics, kinetic and kinematic changes associated with improved concentric performance during stretch shortening cycle movements, compared to isolated concentric contractions. As a result of the dynamics of the stretch shortening cycle, the muscle fibres, during the eccentric phase of contraction, contract in resistance to stretch. This transfers the majority of the stretch to the tendons and tendinous structures. Due to the elastic recoil action of these structures during the subsequent concentric contraction phase, as the force upon them is reduced, the physical shortening (i.e. length change), and velocity of shortening within the muscle-tendon complex, is greater in the tendons (and possibly aponeuroses) than the muscle fibres. This allows the muscle fibres to function at relatively greater lengths, and at slower shortening velocities, than would normally occur at the same movement speed. This improves the length-tension and force-velocity relationship of the active muscle fibres, thereby allowing them to contract more forcefully.

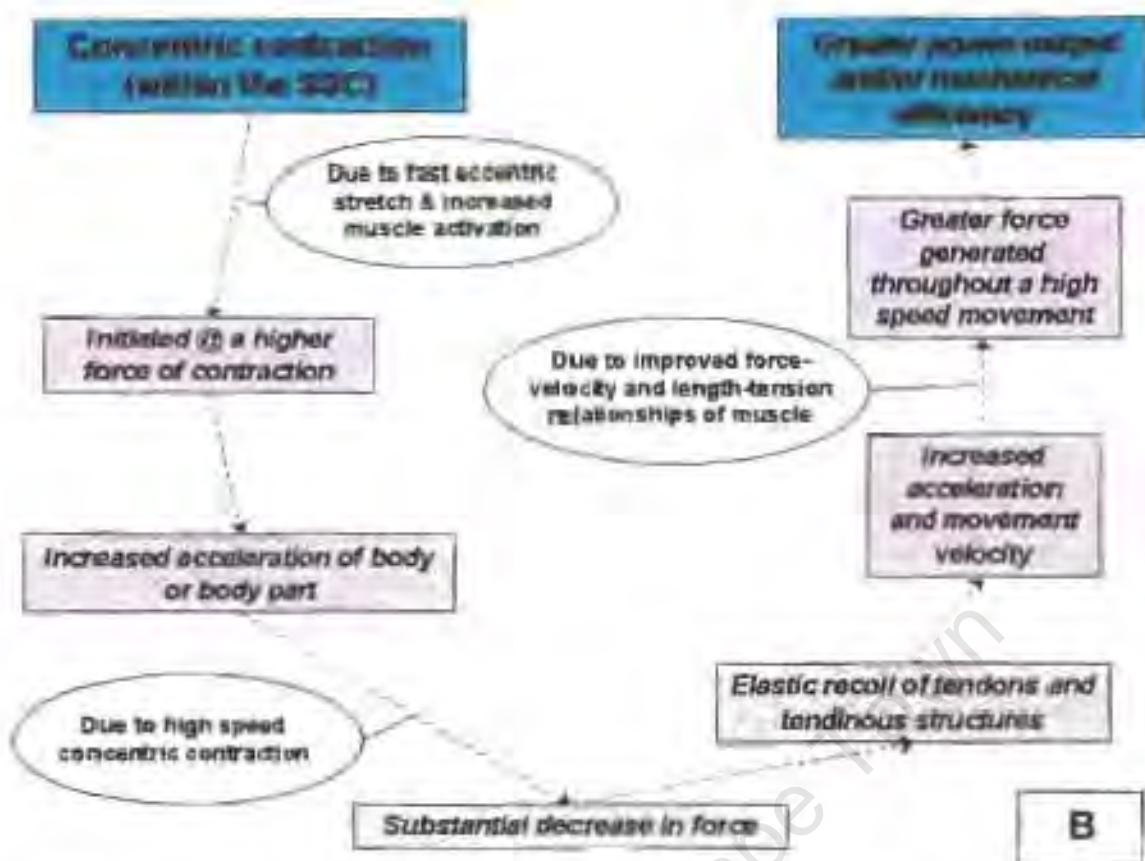
### ***SUMMARY OF STRETCH SHORTENING CYCLE POTENTIATION***

Many practitioners in the field believe that concentric performance enhancement is stretch reflex facilitated, while the majority of the research seems to indicate it is an elastic phenomenon (134). The variation in movement strategies during the measurement of stretch shortening cycle and corresponding different kinetics and kinematics, make it difficult to identify any specific individual mechanism that contributes significantly to the potentiation of performance during stretch shortening cycle movements (50). However, an estimation of the relative contributions of elastic and myoelectric potentiation was made at 72% and 28% respectively (91). This calculation may be flawed, because the storage and utilisation of elastic energy is dependent on myoelectrical potentiation (91). Furthermore, the extent of the interaction between elastic energy and myoelectric potentiation may vary between individuals (30), because neural activation and innervation of muscle can be a control mechanism by which sufficient stiffness is formed in the muscle-tendon complex to preserve the tension on the tendon during movement (61).

Figure 6 summarises the mechanisms involved in the process of the stretch shortening cycle of muscle function. This figure shows that the tendons and tendinous structures act as power amplifiers, by increasing the movement speed, and improving the dynamics of the involved muscle fibres. Therefore, as the tendons play such an integral role in the storage and release of elastic energy, elastic recoil assistance, the altering of muscle-fibre dynamics and force transmission during the stretch shortening cycle (Figure 6), it is logical to assume that variations in elastic stiffness of the tendons could significantly impact on the mechanistic functioning and relative contributions of the involved processes. The next section of this review will discuss musculotendinous stiffness in more detail.



(Legend on next page)



**Figure 6:** A summary of the mechanisms involved in the process of the stretch shortening cycle of muscle function. **A** – represents the process forming the eccentric contraction phase  
**B** – represents the process forming the concentric contraction phase

## MUSCULOTENDINOUS STIFFNESS

A muscle-tendon complex is responsible for the movement of joints and has functional properties. These properties vary depending on the stiffness or elasticity of the complex, and the dynamics of the actively involved muscles (38).

As described earlier, the enhancement of concentric performance by active pre-stretch, during stretch shortening cycle movements, can be attributed in part to the release of elastic energy stored in the series elastic components of muscle (27). The series elastic components of a muscle-tendon complex consist of both passive and active elements of muscle i.e. the tendon, aponeurosis and actin-myosin cross-bridges, and are considered to be the most important part of the muscular elastic elements for the functioning of active muscle (79).



A significant amount of series elasticity resides in the cross-bridges of the muscles themselves (33;46). Cross-bridges also contribute to the force-stiffness relationship of the entire muscle-tendon complex (46). However, the major part of this series elasticity resides in the tendon (46;132;148) and its stiffness is largely dependent on the contractile components of muscle (132).

### ***Elasticity***

The relationship between length and force gives important information concerning the elastic behaviour of a biological material (62). Elastic behaviour is portrayed by the relationship between the change in length of the involved tissue and the respective force applied to it (132). When an element has elastic properties it is deformed, i.e. undergoes a change in length, by a force and regains its initial form after the force has been removed (64). Elasticity is a characteristic property of any substance to resist a change in length and return to its normal form (87). This leads to the concept of stiffness or conversely compliance.

The extensibility of the soft tissue structures within a muscle-tendon unit is related to the resistance they offer during lengthening (112). Stiffness is defined as the resistance of a tissue or material to stretch (87). More specifically, stiffness can be defined as the force required to stretch a tissue or biological material by a specific amount (e.g.  $\text{N}\cdot\text{mm}^{-1}$ ) (33;87). Compliance, the inverse of stiffness, is defined as the ability of the tissue or material to stretch (87). More specifically, compliance describes the amount that a tissue or biological material has been stretched in relation to a specific force applied upon it (e.g.  $\text{mm}\cdot\text{N}^{-1}$ ) (33). A muscle-tendon complex that is more compliant (i.e. less stiff) will elongate more with less resistance (112).

The elasticity of the muscle or muscle-tendon complex, is usually measured either as stiffness, or as compliance (38;98). The stiffness or force-extension curve of the elastic components of the body follows a similar path to that of the series elastic components in isolated muscle (32).

In summary, elasticity contributes to stretch shortening cycle exercise potentiation (9), and stiffness dimensions provide an objective measure of the elasticity of the muscle-tendon complex (111).

# **MEASUREMENT OF MUSCULOTENDINOUS STIFFNESS**

## **INTRODUCTION**

Various testing strategies and procedures have been used to test for stretch shortening cycle function. These have varied from vertical jump (7;13;62;89;151) and bench throw isotonic variations (42;119;153;156), isokinetic testing variations (136-140), and *in vivo* measures (12;56). All methods have related their findings to the elastic properties of muscle, in specific the series elastic components of muscle. Most of the techniques for measuring these components are invasive or designed around biomechanical simulation models (151). There are however a few non-invasive methods that have previously been used in research, to determine the elastic stiffness (alternatively compliance) of the muscle-tendon complex in humans. These techniques have varied from using force-oscillation vibration physics (32;80;112;132;149;151-153;155;156), isokinetic testing (136) to ultrasonography (96-102). The standard for functionally and non-invasively testing for series elastic component elasticity appears to be the oscillation technique.

## **OSCILLATION TECHNIQUE**

One can assess active musculotendinous stiffness by applying a mild perturbation to a loaded muscle-tendon unit and analysing the induced force oscillations (80). The oscillation technique originates from earlier studies by Cavagna (32), in which he measured the stiffness of the elastic structures of muscle. He measured the damped vertical force oscillations of the body after a sudden deceleration of contracted muscle. Damping refers to the reduction in vibration through dissipation of energy (32;106). Damping of vibration is an expected phenomenon as it represents the friction within muscle, the surrounding tissues, the involved joints and the antagonistic muscles (32;152).

Assuming that the oscillations of the muscle-tendon complex under different loading conditions were similar to that of damped harmonic motion, Cavagna (32) calculated the elastic stiffness and damping coefficients of these structures, and also estimated the amount of stored mechanical energy. The damping coefficient is determined by the exponential decrease in the force peaks of successive oscillations. The natural log of the

exponential decay of the peak oscillatory forces over time forms a linear relationship, from which the damping coefficient can be determined (32).

For the measurement of musculotendinous stiffness using this technique, the muscle group is loaded in a semi-static or stationary position, and an abrupt external force or perturbation is applied to the system. For example if a limb were held in a stationary position by springs (e.g. muscle-tendon complex), any disturbance created by the application of a force would create oscillations (126). These fluctuations or oscillations in force registration are used to calculate the stiffness of the muscle-tendon complex, based on the assumption that the characteristics of the oscillations are typical of a damped harmonic motion (32). Stiffness or elasticity is calculated from the frequency of the oscillations and the damping coefficient (32).

Measures of musculotendinous stiffness assume that the spring-like properties of the series elastic components are reflected in the response of the entire system, and that measures of stiffness are the result of the interaction between the separate muscle groups involved (132;151).

During most tests of muscle function multiple joints are incorporated into the test as muscle groups function as synergistic units in combination during sporting movements, and therefore also stretch shortening cycle movements. For example during vertical jumping, which is a multi-joint movement, the resulting jump is a function of the combined impulses from all the participating muscles (25), and not any one specific muscle. It would therefore seem logical to evaluate them functionally. The stiffness measure would thus represent that of the entire system or combination of muscle groups.

Because the oscillation technique measures the elastic property of the entire system, and not individual muscles or muscle groups, it is impossible to isolate the individual contributions of the various series elastic components (151). It must therefore be noted that the stiffness measures of, for example the lower body oscillation test, reflect the combined elastic properties of all the lower body musculature i.e. hamstrings, quadriceps, gluteals and triceps surae muscle groups (151).

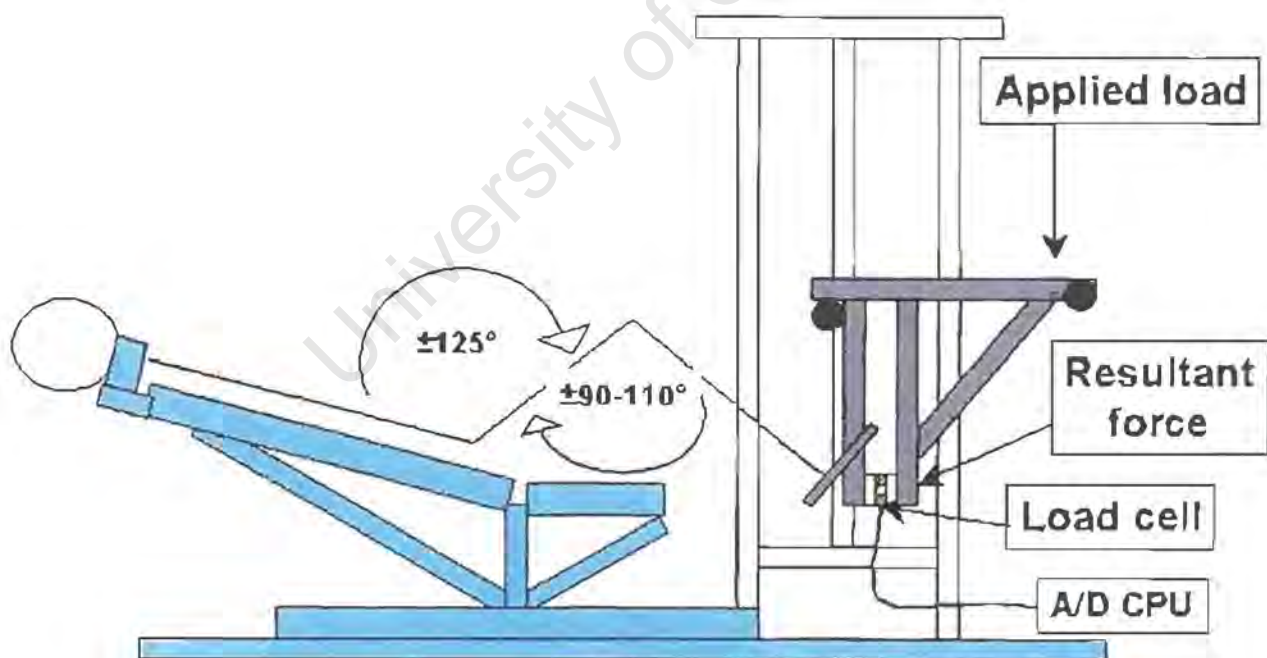


## Lower body

The testing position in the lower body procedure is specific to the jump tests used in the literature. Therefore the overall elastic contribution of the associated elements can be compared (151). Similarly, it is reasonable to assume that the overall force developed in total body movements, e.g. in vertical jumps, relates to the average muscle tendon lengths (159).

### Equipment

To measure musculotendinous stiffness of the lower extremities, this method uses a rigid metal structure that pivots freely on an Olympic bar that is connected to a leg press device, designed specifically for this procedure (Figure 7). The stiffness system (rigid metal structure) is allowed to pivot freely on the Olympic bar. The horizontal bar can be loaded with up to 240 kg, and functions as a pendulum.



**Figure 7:** A schematic representation of the stiffness system and positioning for testing lower body stiffness. Modified from Walshe and Wilson (149).

A load cell and force transducer is attached to this pivoting metal structure, between the two rotating arms of the system. This measures the compressive force or applied tension (70). The compressive force is a direct measure of the force at the subject's feet (151). The load cell is connected via a cable to an analogue-to-digital converter (A/D, Figure 7) and computer (CPU, Figure 7), which records the force outputs.

### **Procedure**

For the oscillation to be measured, the subject's positioning needs to be standardized. A quasi-static isometric leg press position is used, where the subject is instructed to maintain a hip angle of approximately 125°, and a knee angle of approximately 100° for between 1 and 3 seconds. The joint angles are measured with a goniometer.

During this time, the tester briefly perturbs the system, by applying an approximate external force of 100 - 200 Newton to the rigid structure indicated in Figure 7. This procedure of perturbing a loaded muscle-tendon complex has frequently been used in the literature (32;80;132;149;151;152;155;156). Even though this perturbation will undoubtedly vary between subjects and trials, the stiffness values should not be affected, because an elastic system should oscillate at its natural frequency, independent of the magnitude of the perturbation (80;112;149;151-153;155;156).

The natural frequency of oscillation is an important feature of an elastic system. When a perturbation is applied, an elastic system is disturbed from its static equilibrium, and vibrates at its own natural frequency (39;112;132). In the body the damped natural frequency of oscillation has been reported as being within a 1-6 Hz range (32;55;57;65;106;132;156) and is determined by the load on the system as well as its stiffness (132).

The abovementioned system is modelled on a viscous system and is a function of both the natural frequency and damping (112). The damped oscillations fade exponentially depending on the combined viscoelastic properties of the entire system (132). The elastic and damping constants are calculated using the reaction force of the system, generated as a result of the applied perturbation (57). Considering these variables are affected by the mass on the system (112), the resulting damped oscillation to measure elastic stiffness is

expressed by the following equation, which is representative of a damped mass-spring model (132;149;151-153;155;156):

$$m(d^2x/dt^2)+c dx/dt+kx = mg$$

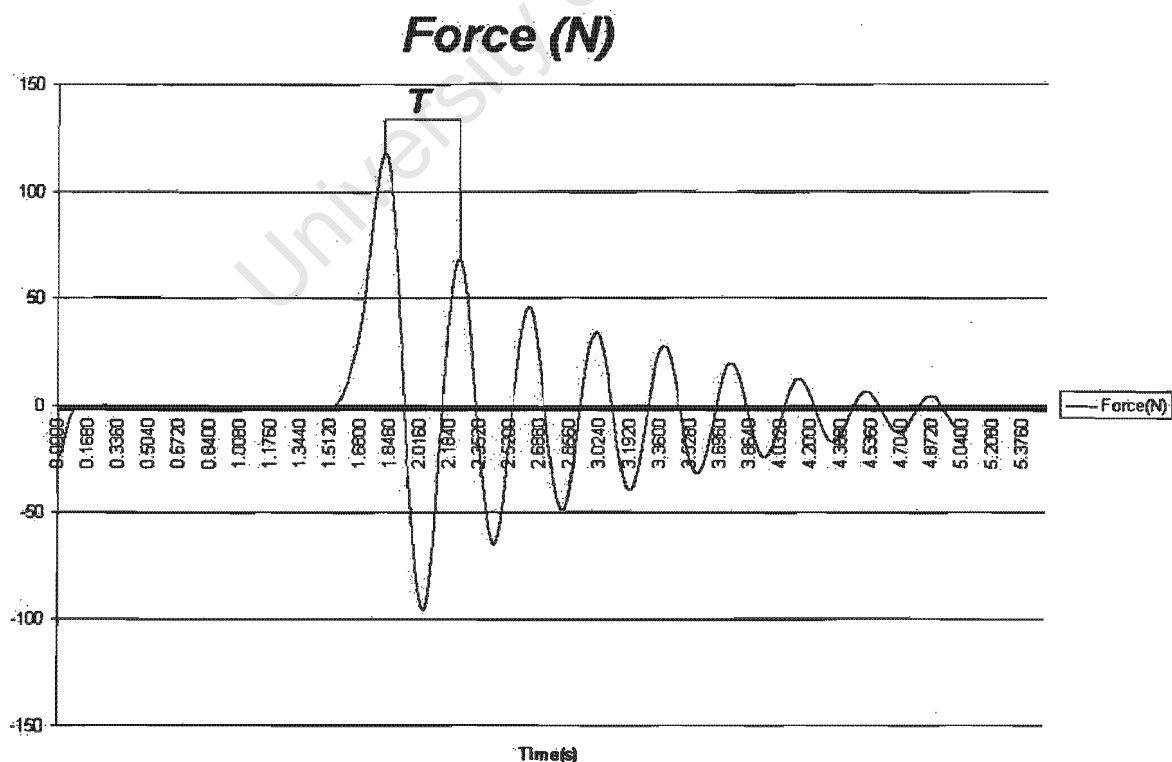
Where:

- $x$  = position (m),
- $g$  = acceleration of gravity ( $m.s^{-2}$ ),
- $m$  = mass of the system and weights (kg) and
- $k$  = stiffness ( $N.m^{-1}$ ).

Stiffness is determined by the equation (32;80;112;149;151;153;155;156):

$$k = 4(mf^2\pi^2)+(c^2/4m)$$

Where  $f$  = damped natural frequency, or the inverse of the time-period between two force peaks ( $1/T$ , Figure 8). After plotting the natural log of peak forces against time, and determining the ratio of change in peak force and its relationship to the damping in vibration, the damping ratio  $s$  is calculated.



**Figure 8:** A schematic representation of a damped force plate oscillation, where  $T$  represents the time-period between peak forces used to calculate damped natural frequency. (\* Graph from original data)

The following equation determines the damping coefficient  $c$  (149;151;153;155;156):

$$c = 4 \pi m f_n s$$

Where natural frequency ( $f_n$ ) was calculated using the equation (149;151;153;155;156):

$$f_n = \{f^2/(1-s^2)\}^{1/2}$$

According to this test each subject is tested at various percentages of maximum isometric leg press, i.e. 30, 40, 50, 60, and 70% of maximum. Figure 8 represents a resulting damped oscillation of a muscle-tendon complex, which represents the viscoelastic properties of human muscle.

### Maximal stiffness

Cavagna (32) validated the oscillation technique for measuring musculotendinous stiffness, and found that at submaximal loads, the stiffness of the muscle-tendon complex increases with increasing force. However, with increasing loads, stiffness gradually forms a plateau (32;46;132;151;153;156). Using the inverse of stiffness, which is compliance, Kubo et al. (97;98;101) found a similar tendency. Using a different methodology, Kubo et al. (97;98;101) showed that the compliance of the muscle-tendon complex decreased, and almost became constant in the 50 - 100% range of maximal voluntary contraction. They used the averaged value of the change in length, divided by the change in force above the 50% load, as a measure of the compliance of the tendon structures.

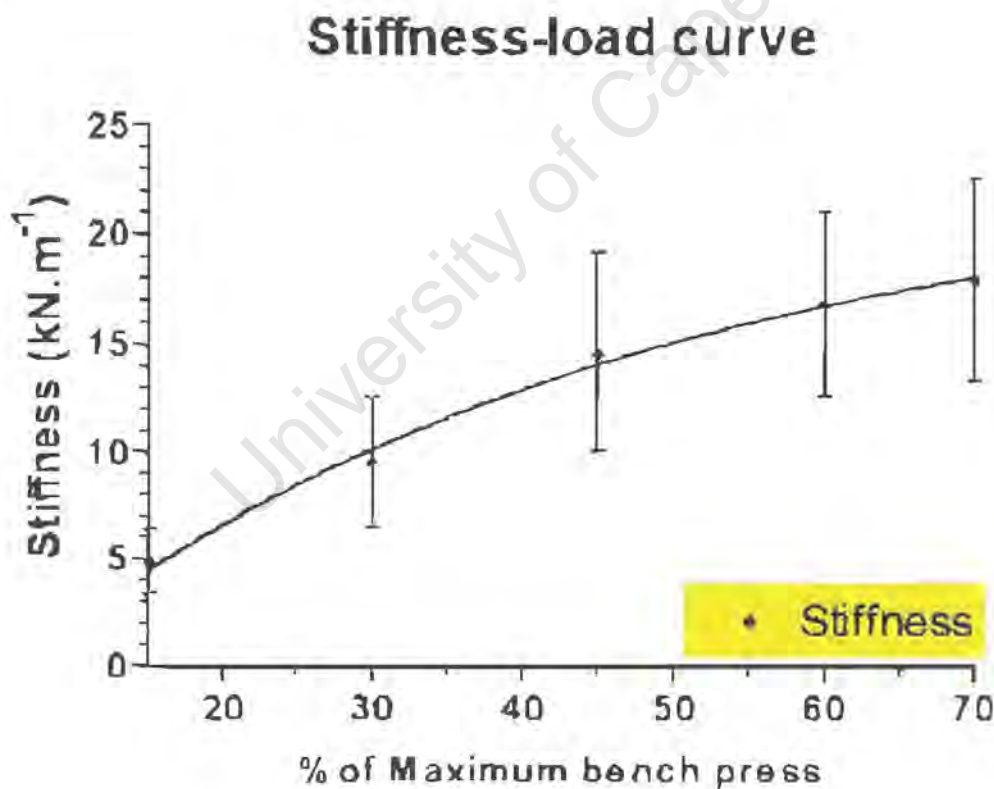
The measurement of maximal stiffness is important, as it can be inferred that the stiffness of the muscle approaches the stiffness of the tendon, as the load on the muscle-tendon complex increases (38;115;132;153). Shorten (132) states that muscle stiffness appears to be the main contributor to musculotendinous stiffness at low loads, whereas the more compliant tendon becomes the main contributor as the load increases. This implies that with incremental loading the muscle stiffness increases and eventually equals or exceeds that of the tendon, and that at high loads, the stiffness of the system or muscle-tendon complex therefore represents the stiffness of the more compliant tendon (124).

According to Cook and McDonagh (38), at full activation, which would be equivalent to maximal loading, the stiffness of muscle and tendon would be equal (115). Hence, the use of the oscillation technique for determination of maximal stiffness can be seen as a representative measure of the average stiffness of the tendons of the involved muscle groups.

Maximal musculotendinous stiffness, using the oscillation technique, is determined by the use of a downward exponential extrapolation, by plotting the average stiffness per load i.e. at 30, 40, 50, 60 and 70% of maximum and fitting it to the equation (149,151):

$$k = f(1) = a x e^{1/b} + c$$

Where **a**, **b** and **c** are constants and the specific curve fit is determined by the least sum of squares criterion (Figure 9) (149;151).



**Figure 9:** A graphic representation of the average stiffness at each load  $\pm$  SD, represented by data points and the vertical bars respectively. The line of best fit used for calculating maximal stiffness, is also shown, using a downward exponential association and the least sum of squares criterion. Modified from Wilson et al. (156).

This technique of measurement of musculotendinous stiffness has been used in a variety of research publications (149;151;153;155;156).

### **Reliability and validity**

This *in vivo* measure of musculotendinous stiffness is performed in a functional, closed chain movement similar to that found in running and jumping (151). This technique of measuring musculotendinous stiffness in the lower body is both valid and reliable with a day-to-day reliability of  $r = 0.94$  and a coefficient of variation of 8% (151).

### ***Upper body***

#### **Equipment**

A comparable procedure to measure musculotendinous stiffness of the upper body musculature has also been used in previous research (153;155;156). The major muscle groups used for these experiments were the chest and upper limb musculature in the form of a bench press movement.

The test is performed on a bench conforming to international standards for the performance of a bench press movement. The flat bench bolts onto a force platform, to limit horizontal movement and direct forces in one area (153;155;156). This procedure improves the sensitivity of the apparatus to force fluctuations. It also allows isolation of the vertical ground reaction force, or the vertical force exerted by the subject against the bar (119;154).

#### **Procedure**

The hand position on the bar is standardized as the distance between elbows with the arms abducted 90° (154). The subject is instructed to maintain the bar in a quasi-static position ~3 cm above the chest. The subject has to maintain this position for ~0.6 s, after which the tester briefly perturbs the system and pushes down on the middle of the bar with an external force of ~100 - 200 N (156). The tester instructs subjects not to respond to the perturbation of the bar to limit neural response (153;155;156). The same calculation procedure described for the lower body tests (p 62 - 64) is used for calculating musculotendinous stiffness of the upper body.



### **Maximal stiffness**

The loads tested for the upper body are similar to the relative loads of the lower body, i.e. 15, 30, 45, 60 and 70% of maximal (1RM) rebound bench press (RBP) (153;156). Each test load is repeated three times for each subject and the values at each load averaged.

Note that in both cases, i.e. the upper and lower body, the heaviest load was only 70% of maximal load. It has been established that loads above 70% of maximal load are too heavy, and cannot be maintained with sufficient stability for the duration of the oscillations (156). The authors also found that the variability of the stiffness data increased as the load increased. With this in mind, they stated that it was more reliable to predict maximal stiffness using the curvilinear stiffness-load relationship than trying to measure it directly (156).

### **Reliability and validity**

The day-to-day reliability of this technique in the upper body was  $r = 0.89$  (153).

### **Electromyography (EMG)**

In all the *in vivo* techniques for measuring musculotendinous stiffness, a problem that needs attention is that of preactivation. Previous protocols have used EMG, as a control measure to eliminate confounding due to preactivation in the oscillation technique (149;151;156). Surface EMG is recorded 150 ms before and 150 ms after the perturbation to determine the effect of preactivation and to negate the effect of reflex potentiation on musculotendinous stiffness (149). This time-period (150 ms) has been used as reference, because Zhou et al. (160) established that it was within this time-period that any reflex contributions would be observed, i.e. increased myoelectrical activity.

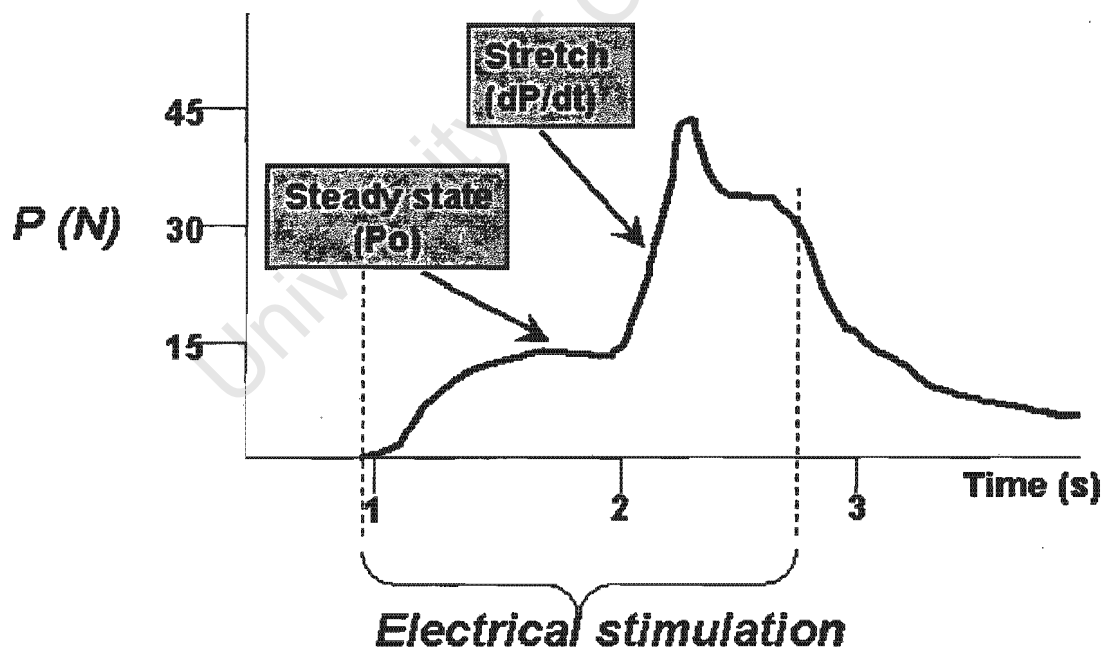
### **ISOKINETIC TESTING**

Previous research has also used an isokinetic dynamometer to measure musculotendinous stiffness (136). Svantesson et al. (136) used an isokinetic

dynamometer (Kin-Com II, Chattanooga Group, Inc., Hixson, TN, U.S.A.) to isolate the calf muscles, and to measure musculotendinous stiffness.

### ***Procedure***

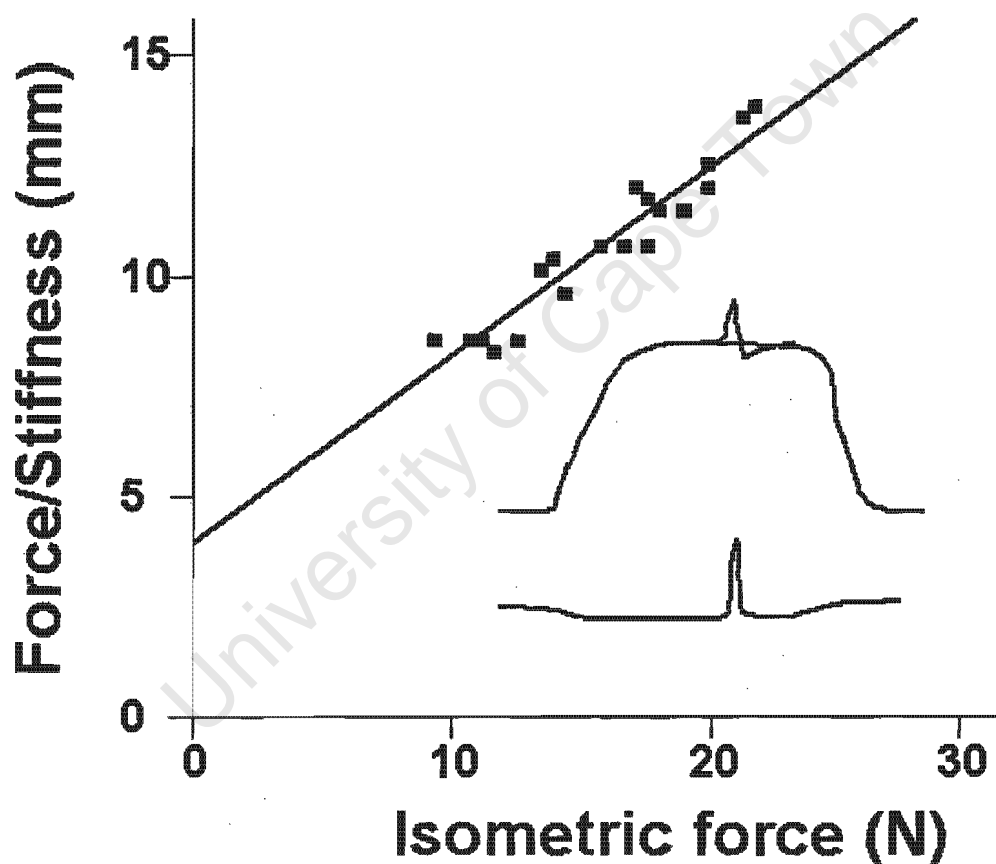
The starting position of the ankle joint is in  $10^\circ$  of plantar flexion, with the subject prone-lying with hips and knees fully extended. Using this particular technique, electro-stimulation is incorporated, where the triceps surae muscle group is pre-stimulated until a tetanic contraction of  $\sim 15 - 30\%$  of maximal force output (MVC) is reached. After reaching tetanic state, a short stretch is applied moving at  $200^\circ \cdot s^{-1}$ , from  $10^\circ$  plantar flexion to  $10^\circ$  dorsiflexion, a total movement range of  $20^\circ$ . The effect of this action is recorded and immediately after the stretch, the electro-stimulation is terminated (Figure 10).



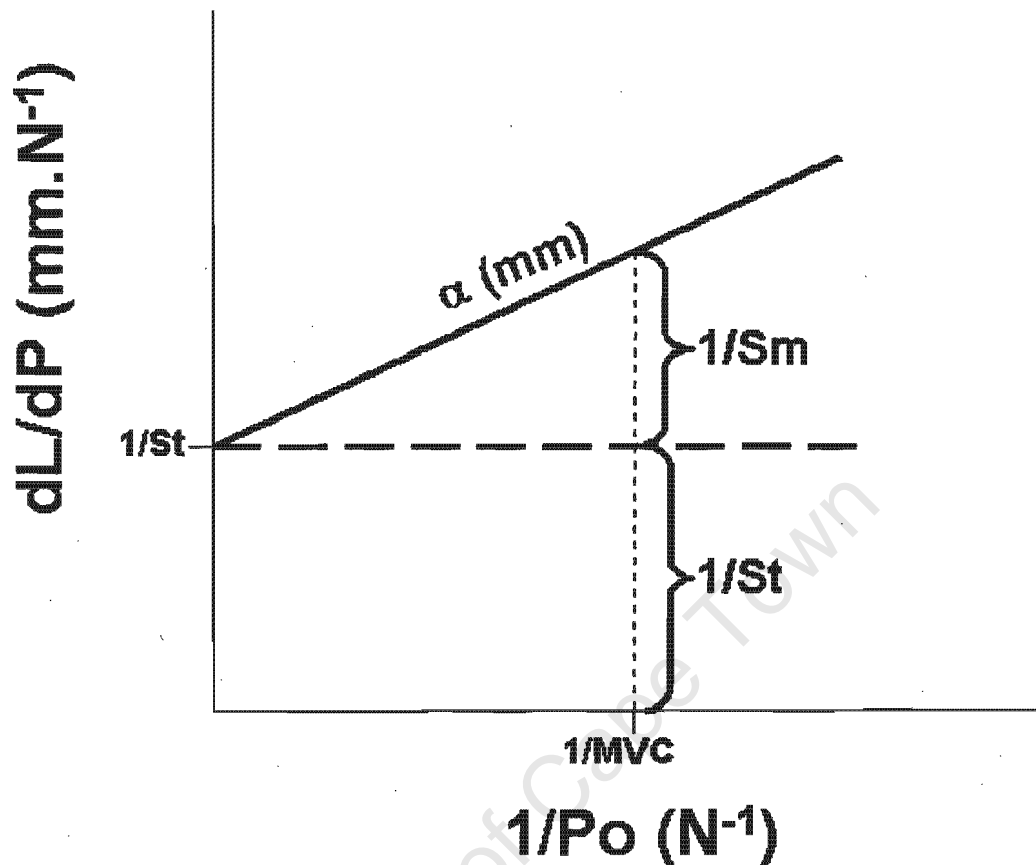
**Figure 10:** The change in force ( $dP/dt$ ), when a stretch has been applied to an electrically stimulated tetanic contraction ( $P_o$ ). Modified from Svantesson et al. (136).



For the stiffness test, the force curves are analysed using the methods of Cook and McDonagh (Figure 11) (38;136). This technique differentiates between tendon stiffness ( $S_t$ ) and muscle stiffness ( $S_m$ ) and defines total stiffness, as the inverse sum of the inverted muscle stiffness and inverted tendon stiffness (Figure 12). The technique is a modification of the original technique of Morgan (115), which also differentiates between tendon and muscle stiffness (38;136).



**Figure 11:** The isometric force/total stiffness (y-axis) is plotted against the isometric force (x-axis) and the line of best fit obtained for the subject values. The gradient of this line is the reciprocal of tendon stiffness and the y-axis intercept ( $\alpha$ ) is used to calculate muscle stiffness at each load. Modified from Cook and McDonagh (38).



**Figure 12:** The separation of muscle- ( $S_m$ ) and tendon ( $S_t$ ) stiffness components using the total compliance (inverse of stiffness) ( $dL/dP$ ) at each level of electrical stimulation induced by stretching. Modified from Svantesson et al. (136).

Muscle and tendon are arranged longitudinally or in series, therefore the total stiffness of the muscle-tendon complex is the function of the sum of tendon stiffness and muscle stiffness (38).

Most techniques for differentiation of tendon- and muscle stiffness involve dissection, and are therefore not suitable for humans (38). The method used by Cook and McDonagh (38) does not involve dissection or removal of body tissue. The muscle is stimulated and after tetarisation, a small stretch is applied to the muscle, and the change in force is registered via the transducer, in this case the isokinetic dynamometer.

The total stiffness (inverse of total compliance) of the muscle-tendon complex is calculated, as the difference between the tetanic force and the force induced by the stretch, divided by the change in length of the muscle-tendon complex (38).

$$\text{Total stiffness} = dP/dl$$

Tendon stiffness above 25% tension essentially remains constant (124). After a tendon is stretched ~4% of its resting length and tension has reached approximately one third of its maximum isometric value, the stiffness of isolated tendon becomes constant (124). In other words, once this initial tension has been taken up by the system, the length-tension relationship of the tendon becomes more linear (124).

Tendon stiffness is therefore defined as the constant  $k$  and muscle stiffness ( $M$ ) is defined as:

$$M = \text{force}/\alpha$$

Where  $\alpha$  is a constant with the dimension of length.

Using this premise, the total stiffness is calculated from (38;136):

$$1/\text{total stiffness} = 1/\text{tendon stiffness} + 1/\text{muscle stiffness}$$

or

$$1/\text{total stiffness} = 1/k + \alpha/\text{force}$$

The change in length of the muscle-tendon complex was calculated as 6 mm during the 0.1s period of active stretch (136). The force difference between tetanic stimulation and stretch is measured and total stiffness ( $dP/dl$ ) is calculated. This procedure is repeated several times using different voltage levels ranging between 15 and 30% MVC (136).

Total compliance ( $dl/dP$ ) ( $1/\text{total stiffness}$ ) data are then plotted against the inverse of the force ( $1/dP$ ) data over the range of voltage levels tested (Figure 12). A simple regression equation is used and the calculated intercept is equivalent to the inverse of the tendon stiffness ( $1/\text{tendon stiffness}$  or  $1/k$ ) and the slope, equivalent to the constant  $\alpha$  (136).

Maximal muscle stiffness is then calculated for maximal isometric force (MVC) and total stiffness is then calculated using the abovementioned formulae.

Alternatively to make this formula more usable, Cook and McDonagh (38) multiplied the equation by force, which modifies the equation to:

$$\text{Force/total stiffness} = \text{force}/k + \alpha$$

The resultant force data ( $dP$ ) are then divided by their respective total muscle-tendon complex stiffness values (**total stiffness**), and these calculated values (**force/total stiffness**) are then plotted on a graph against the corresponding isometric force (Figure 11). A linear force- $\alpha$  curve is generated, where  $\alpha$  is the force/total stiffness ratio.

To fit a straight-line relationship to these data, the method of least squares is used. The resultant regression equation is then used to determine tendon stiffness and muscle stiffness at any corresponding isometric force (38). The gradient of this line is the reciprocal of tendon stiffness ( $1/k$ ) and the Y-intercept is the constant  $\alpha$ . From this, one can separate muscle and tendon stiffness from the formulae above (38), as well as calculate maximal muscle ( $MVC/\alpha$ ) and total stiffness (136).

Maganaris and Paul (110) however criticised the use of this technique as it assumes a linear force-lengthening relationship of the muscle-tendon complex despite evidence to the contrary.

#### ULTRASOUND TECHNIQUES

Kubo et al. (96-102) used ultrasound as a technique for quantifying the elastic stiffness of the tendons of various muscles *in vivo*. This technique used real-time ultrasonography to measure muscle-tendon complex extensibility, or conversely stiffness, by measuring the elongation of the tendon and aponeurosis of muscles during isometric contraction (96-102). Kubo et al. (97;98;101) obtained an ultrasonic image from the vastus lateralis at the mid-thigh position (halfway between the greater trochanter and lateral epicondyle of the knee). To standardise their procedure, and ensure that the probe did not move during the test, they place an anti-radioactive marker on the skin surface. An echo was sent from

the apparatus at this marker (97;98;101). From the ultrasonic image, they could differentiate between the vastus lateralis and vastus intermedius muscles and the deep aponeurosis that separates them. A cross-point from two echoes where a fascicle from vastus lateralis attached itself on the deep aponeurosis was sourced (97;98;101).

As the point on the superficial aponeurosis did not move during the procedure, they then assumed that any proximal displacement of this cross-point would represent the stretching of the deep aponeurosis and distal tendon (97;98;101). Isometric muscle shortening is only possible if the series elastic elements at the ends of the working muscle (i.e. tendons) are lengthened or stretched (128). Hence any shortening of the engaged muscle fascicles would represent the lengthening of the tendon, and from this one could calculate the compliance/stiffness of the tendons (128). Isometric force data were collected using a dynamometer with the subjects seated upright with both hip and knee in 80° flexion (97;98;101). Ito et al. (84) also used a similar procedure using the ankle dorsi-flexor, tibialis anterior muscle, measured in 20° of plantar flexion. Kubo et al. (99;100;102) used a similar procedure on the lower leg using the medial gastrocnemius muscle.

Firstly maximal voluntary contraction force (**MVC**) was determined in this position. Following this the change in length of the distal tendon and deep aponeurosis was measured in increments of 10% of the MVC until maximum was obtained (97;98;101). Force was calculated from the generated torque using the equation (97;98):

$$F = k.\tau.MA^{-1}$$

Where:

- **k** = Relative force contribution
- $\tau$  = Torque generated
- **MA** = Length of the moment arm at 80° flexion

The relative force contribution of the vastus lateralis muscle was calculated as 22% of the total knee extension torque. This was based on the vastus lateralis muscles' percentage of the cross-sectional area of the quadriceps femoris muscle group (97;98;101).

The stiffness value (**N/mm**) (101) or compliance value (**mm/N**) (97;98) for each load was then plotted for every 10% increment, and a curvilinear relationship was shown. The

stiffness-load curve initially increased almost linearly, but after  $\pm 50\%$  the increments in stiffness no longer became significant and the curve started to plateau (97;98;101). Kubo et al. (97;98;101) performed a linear regression using the stiffness data from 50 - 100% of MVC and obtained the tendon stiffness value from the slope. Their rationale for using the data points 50 - 100% was that the values subsequent to 50% of MVC did not significantly differ from one another (97;98;101).

The stiffer the tendon, the less distance the cross-point moves towards the anti-radioactive marker, because of the increased resistance against elongation supplied by the stiffer tendon, and vice versa (97;98;101). This technique would however be difficult to use for combined muscle groups and would also be extremely time consuming and impractical.

Maganaris (109;110) also used ultrasonography for measuring musculotendinous stiffness. He used two different muscle groups to compare and relate their differences in stiffness profiles. A similar yet more controlled procedure to the abovementioned techniques was used to determine the elongation of the respective tendons. Two superficial muscles namely, tibialis anterior (109;110) and gastrocnemius (109) were used, as they allowed for high quality ultrasound measurement. To eliminate the effects of co-activation of antagonistic muscles, the agonist muscles were electrically stimulated to contract and their force-length data were therefore seen as true and representative of the muscles tested only (109;110). To avoid artificial elongation of the tendons, the ankle was strapped into position on the isokinetic dynamometer with inelastic tape. Other internal controls were made for slippage or movement of the respective muscles' tendon origins and insertions. The tibialis anterior tendon had a maximal elongation of 2.5% strain and the gastrocnemius tendon had a maximal elongation of 4.9% strain. The stiffness values of the tibialis anterior tendon and gastrocnemius tendons were  $150 \text{ N.mm}^{-1}$  and  $160 \text{ N.mm}^{-1}$  respectively. These differences in compliance were attributed to their different functionality in the body.

Muramatsu et al. (118) also used ultrasonography to measure the *in vivo* elastic characteristics of the medial gastrocnemius muscle and its tendinous structures i.e. achilles tendon and the proximal and distal aponeuroses. To evaluate the movement of these structures independently, they used an ultrasonic probe placed at the myotendinous junction, at the proximal intersection of the muscle fascicles and the aponeurosis, and

another midway in between these points. The achilles tendon length was determined from an ultrasound scan as the distance between the probe at the myotendinous junction and the osteotendinous junction at the calcaneus. Using a similar incremental loading procedure to the abovementioned studies, they showed no significant difference in strain between these structures. The achilles elongated  $5.1 \pm 1.1\%$  and the aponeurosis  $5.9 \pm 1.6\%$  at maximum. Muramatsu et al. (118) proposed that the lack of difference between these two structures might be ascribed to the difference in cross sectional area. They further showed that the strain of the tendinous structures i.e. both tendon and aponeurosis, did not increase significantly above 30 and 60% of MVC respectively.

Griffiths (67) also used ultrasonography to determine muscle fibre shortening during active stretch, using muscle tissue from cats. Piezoelectric crystals were implanted into the ends of a bundle of muscle fibres. Fibre length was calculated by pulsing ultrasound and recording the delay in sound wave propagation between the two piezoelectric crystals. This technique has not been used in human tissue due to the invasiveness of the procedure.

### **STIFFNESS EXPLAINED BY AN ELASTIC MODULUS**

Muscles and tendons have been described as having spring-like properties (44;132;151). High muscle stiffness, together with ideal spring-like properties, however, would make it impossible to absorb the impact and remain on the ground after a jump (44). The regulation of muscle stiffness, together with the spring-like properties can influence the ability of these muscles to store and mobilise energy (44). Both tendon and muscle have mechanical properties that can be explained by relatively basic elastic models (57;132).

In the field of physics, there is complex relationship between stress, strain and stiffness as explained by Young's modulus (31;124). Stress is defined as the force applied per unit area and is measured in  $\text{N.m}^{-2}$  or Pascal (Pa). Any stress applied to an object, for example a muscle-tendon complex, will cause strain, or extension of the muscle-tendon complex (the extent dependent on the stiffness of these tissues) (31), thereby increasing the amount of strain (elastic) energy stored (132). Strain is actively influenced by the stress applied to the system.

The measure of stiffness, or resistance to stretch, could therefore be defined by an elastic modulus (132). In the example below, Young's modulus of elasticity is expressed by the following equation:

$$\begin{aligned}\text{Young's Modulus (Y)} &= \text{stress/strain} \\ \text{OR} \\ &= (F/A) / (\Delta L/L)\end{aligned}$$

Where:

- F = force,
- A = cross-sectional area,
- $\Delta L$  = change in length, and
- L = resting length.

Stiffness essentially means the ratio between the change in force (F) and the change in length ( $\Delta L$ ) i.e.  $(F/\Delta L)$  (46). A larger stiffness-value would therefore indicate that more force is needed per unit area to stretch the muscle-tendon complex, indicating increased stiffness or decreased compliance of the muscle-tendon complex. Essentially a modulus of elasticity is a measure of the stiffness of a structure (131).

## **MUSCLE STIFFNESS**

Muscle stiffness is primarily regulated by electrical activation of the involved muscles (61) and in the context of movements involving the stretch shortening cycle, is mainly related to specific neural mechanisms (33). These mechanisms include stretch reflex activation during a fast eccentric stretch, or central preactivation of muscle before ground or resistance contact. Both mechanisms increase muscle stiffness or resistance to stretch during impact loading.

When a muscle membrane or sarcolemmal potential is generated, the  $\text{Ca}^{2+}$ -ions diffuse from the sarcoplasmic reticulum into the sarcoplasm, bind with troponin and cause cross-bridge interaction (81). To restore the muscle to its resting state, the  $\text{Ca}^{2+}$  is actively



transported into the longitudinal elements of the sarcoplasmic reticulum, thereby reducing the cytosolic  $\text{Ca}^{2+}$  concentration (81).  $\text{Ca}^{2+}$  is removed from troponin, which causes detachment of the actomyosin cross-bridge and relaxation of the muscle. This balance between relaxation and activation determines muscle stiffness.

When increased central drive occurs, for example in the preactivation phase of stretch shortening cycle movements (62), the balance between release of  $\text{Ca}^{2+}$  from the sarcoplasmic reticulum and removal by the reticular network is shifted towards release (15), and more myosin-heads are kept in the strong-binding position, thereby increasing muscle stiffness (53).

With increasing load, additional fibres are recruited (38;65;119), with corresponding increased  $\text{Ca}^{2+}$ -driven cross-bridge formation, thereby enhancing muscle stiffness (38). Muscle stiffness increases when the muscle is stimulated to contract, as more cross-bridges are formed (38). Increased activation of the K-motorneurons effectively leads to increased stiffness of the involved musculature (121). Stiffness can therefore also be used as a measure of the number of attached cross-bridges (53). Since both muscle force and muscle stiffness are reliant on the number of cross-bridges formed, muscle stiffness increases correspondingly with increasing muscle load (32;115).

Greene and McMahon (65) propose that the increase in the number of activated muscle fibres effectively implies that more springs are added in parallel, thereby increasing the overall muscle stiffness. In other words, muscle fibres are expected to have a stiffness that is proportional to the load carried or muscle tension generated (65), since both variables are dependent on the number of cross-bridges formed (57;124). It has been shown that muscle stiffness increases with muscle force even at maximal isometric loads (46).

## **TENDON STIFFNESS**

Tendon stiffness is normally constant at higher forces (38;124), and a tendon is generally more compliant than the active muscle fibres (67;124;137). Muscle stiffness at higher forces is very high, with the majority of the stretch being transferred to the tendons (124). The tendon therefore stretches more than the active muscle fibres (47;67), and stores the greatest proportion of elastic strain energy (124;137). Since the tendons and tendinous structures are passive structures with the most stretch, they have a potentially important role in muscle function, as they are metabolically efficient, i.e. store and liberate mechanical energy (46). The amount of energy supplied to the skeleton by the elastic tissues in a muscle-tendon complex is largely determined by tendon compliance (4).

Gollhofer et al. (61) also concluded that the majority of elastic energy storage occurred in the tendons. They further reasoned, however, that the nervous system acts as the main "control mechanism" that produces sufficient stiffness in the muscles to transfer the tension or force to the tendons. In addition to tendons functioning as elastic springs, they also function as a mechanical buffer to protect the muscle from rapid length changes (111) and eccentric muscle damage (67;127).

Therefore measuring tendon stiffness could provide insight into muscle functioning, especially during stretch shortening cycle activities. In stretch shortening cycle activities the muscles are eccentrically loaded during impact or countermovement and generate high forces, with the majority of the stretch and tension transferred to the tendon and tendinous structures. Because of the "in-series" design of the muscle-tendon complex, the power, force-velocity dynamics and length changes of the muscle fibres during contraction, in some way, has to be influenced by the elasticity of the tendons (128).

## **MUSCULOTENDINOUS STIFFNESS**

As discussed previously, the compliance or extensibility of the muscle-tendon complex could influence subsequent muscular performance, especially during stretch shortening cycle activities (149). Furthermore, the varied potentiation arising from stretch shortening

cycle exercises may be attributed to individual differences in the stiffness of muscle and tendon (136).

In research, the elastic roles of muscle and tendon are generally viewed separately, yet they function together as a unit during human movement (128). Within a muscle-tendon complex, muscle and tendon are connected in series, and therefore the stiffness of the complex is a representation of the combined stiffness of both muscle and tendon (38;46;124). If one regards both muscle and tendon as individual elastic structures connected in series, any imposed movement on the muscle-tendon complex would affect each subcomponent, i.e. muscle or tendon, differently depending on their respective stiffness values (125). However, the combined stiffness of the muscle-tendon complex is mainly affected by altering the stiffness of muscle and this stiffness is proportional to the force generated (38;136).

Passive stretching of relaxed muscle has no effect on lengthening of the series elastic elements, as the muscle does not resist the stretch with any appreciable force (33;64). Additionally, passive stretching does not represent the structural interaction within the muscle-tendon unit during dynamic functional activity (80). Loading the muscle, however, can significantly influence total musculotendinous stiffness and the lengthening of the series elastic elements. The amount of stretch incurred in the respective components (i.e. muscle and tendon) depends on their respective stiffness values, and the force generated in the muscle-tendon complex (124).

Ettema and Huijing (46) have shown that cross-bridges can have a significant effect on the stiffness of a muscle-tendon complex. During submaximal loading of the muscle-tendon complex, stiffness is generally low because of the lesser muscle stiffness (38). As previously mentioned, with increased loading of the muscle-tendon complex, muscle stiffness will increase, and therefore also its resistance to stretch (38). Because the active muscle resists stretching with such great force during stretch shortening cycle movements, it is this force applied at the extremities of the involved muscle, that facilitates the storage of elastic energy in the series elastic components (33;64). The exact amount of energy stored is, in turn, determined by the stiffness of these structures (64).

## STUDIES RELEVANT TO THESIS

Because of the difficulty in the non-invasive measurement of musculotendinous stiffness *in vivo*, various procedures have been used in an attempt to quantify the elastic properties of muscle and tendon. Most of these procedures represented below (Table 3) measure isolated single joint muscles and compare tendon elasticity to multi-joint movements. These studies therefore do not necessarily present an accurate representation of the true elastic properties of the combination of muscle-tendon units involved in the functional performance tests. To isolate any single muscle as the sole contributor to a functional movement such as vertical jumping or bench press would be inappropriate. One would assume that muscles have a distinct functionality depending on their location in the body and therefore would expect each muscle to have different structural and elastic properties. The techniques of Walshe et al. (149;151) in the lower body and Wilson et al. (153;155) in the upper body seem to be the most applicable, as they incorporate multi-joint procedures and compare the average musculotendinous stiffness of all the involved muscle-tendon units to the various multi-joint functional performance tests. Nevertheless, in most studies, stiffness of the series elastic components of the muscle-tendon complex seems to have an influence on stretch shortening cycle ability. However, this influence appears to be variable depending on the measure and the technique utilised (Table 3).

**Table 3:** Studies relevant to the thesis involving musculotendinous stiffness and stretch shortening cycle in the lower and upper limb musculature.

<b>Reference</b>	<b>Technique</b>	<b>Measured Variable</b>	<b>Muscle's used</b>	<b>Related Variable</b>	<b>Relationship</b>
Kubo et al. (97)	Ultrasound	Tendon compliance	Vastus lateralis	SSC jump potentiation	$r = 0.55$
Walshe et al. (151)	Oscillation	Musculotendinous stiffness	Combined lower limb	SSC jump potentiation SJ height CMJ height RFD SJ RFD CMJ	$r = -0.50$ NR NR $r = 0.54$ NR
Kubo et al. (101)	Ultrasound	Tendon stiffness	Vastus lateralis	SSC jump potentiation SJ height CMJ height	$r = -0.46$ NR NR
Watsford et al. (152)	Oscillation	Musculotendinous stiffness	Triceps surae	SSC jump potentiation RFD	$r = -0.65$ $r = 0.66$
Sikand and Bach (133)	Oscillation	Normalised musculotendinous stiffness	Triceps surae	SJ height CMJ height	$r = 0.61$ $r = 0.55$
Kubo et al. (101)	Ultrasound	Tendon stiffness	Vastus lateralis	SJ height CMJ height	NR NR
Svantesson et al. (136)	Isokinetic	Muscle stiffness Tendon stiffness	Triceps surae Triceps surae	SSC potentiation SSC potentiation	WC* WC*
Wilson et al. (153)	Oscillation	Musculotendinous stiffness	Combined upper limb	SSC impulse potentiation	$r = -0.72$
Wilson et al. (155)	Oscillation	Musculotendinous stiffness	Combined upper limb	Concentric RFD Isometric RFD	$r = 0.65$ $r = 0.78$

**Index:** SSC = stretch shortening cycle, SJ = squat jump, CMJ = countermovement jump, RFD = rate of force development, NR = no relationship, WC\* = weak positive correlation

## **MUSCULOTENDINOUS STIFFNESS AND THE STRETCH SHORTENING CYCLE**

### **TENDON COMPLIANCE**

Strain is the fractional change in length, as a ratio of resting length (31). During contraction, the forces generated by the muscle fibres stretch and tension the more compliant tendon and aponeurosis, before they are transmitted to the bones to produce the torque necessary to create movement (84;85;102;103)}. With the same eccentric force applied on the muscle-tendon complex, a more compliant system will extend more, thereby increasing the strain. Hence the force will act over a longer distance, do more

work, store more elastic (strain) energy (32;132;151;153), and possibly allow for greater stretch shortening cycle performance (151).

Following an intensive stretching programme, tendon compliance increased (153). Wilson et al. (153) also showed that the increased tendon compliance improved elastic energy storage and utilisation. The experimental subjects, who increased their tendon compliance through flexibility training, improved the work performed over the initial 0.37 s of a rebound bench press lift by 20%, which allowed a 5% increase in the load raised (153). Smaller insignificant improvements were noted in the pure concentric bench press lifts in these subjects. The control group, however, remained unchanged in both lifts (153). They therefore attributed the significant stretch shortening cycle performance enhancement, to a reduction in maximal musculotendinous (tendon) stiffness achieved via flexibility training (153).

Considering musculotendinous compliance had previously been shown to contribute positively to stretch shortening cycle activities, Kubo et al. (98) hypothesised that sprinters would have more compliant or extensible tendons in their lower limbs than their matched controls. They therefore tested this theory using an ultrasonic apparatus on the vastus lateralis and medial gastrocnemius muscles during isometric contractions (98). Their data, however, showed no significant difference between the sprinters and controls in the elasticity of their tendon structures in either of the muscle-tendon complexes tested. They did show however, that there was a negative correlation between tendon compliance, in the vastus lateralis and sprint time amongst the sprinters, which may indicate that a more compliant tendon might be advantageous for improving sprint performance (98). They concluded that because of the more compliant tendon structures, and resulting increased elastic contribution, the knee extensor muscle group could generate higher forces during the stance phase of sprinting.

Wilson et al. (156) also found that the maximum use of elastic (strain) energy in a stretch shortening cycle movement occurred with a more flexible or compliant muscle-tendon complex. With eccentric pre-stretch, the more compliant subjects released an additional 37% of energy during the first 0.37 seconds of the concentric contraction compared to the 8% for stiffer subjects. This suggests that more compliant subjects, have a significant advantage over stiffer subjects in stretch shortening cycle movements, and that stretch

shortening cycle muscle performance is possibly a function of musculotendinous stiffness (156).

In a muscle-tendon complex with a more compliant tendon, the more extensible tendon can accommodate much of the stretch, limit myofibrillar displacement (111) and detachment of strained cross-bridges, and thereby enhance the elastic recoil contribution in the ensuing concentric contraction (124). Additionally, the force of the subsequent concentric contraction would therefore be potentiated through improved length-tension and force-velocity relations of the involved muscle fibres (101;149;155;156). Bobbert et al. (19) suggest that without a compliant tendon, one would never be able to achieve the combined high velocities and force-outputs measured.

Kubo et al. (101), after finding no relationship between tendon stiffness and any of their measures (Table 3), subdivided their subjects into a compliant group and a stiff group, and still found no difference between the groups. They did however find a significant difference in the percentage vertical jump height potentiation between the 'compliant' ( $14 \pm 6\%$ ) compared to the 'stiff' group ( $8 \pm 3\%$ ). They further found a significant negative relationship between stiffness and vertical jump height potentiation ( $r = -0.46$ ) (Table 3), and concluded that subjects with more compliant tendinous structures might benefit more from stretch shortening cycle muscle actions.

Kubo et al. (97) showed a significant relationship between tendon compliance of the vastus lateralis muscle and pre-stretch augmentation in vertical jump performance (Table 3). They also found significantly lower jump heights and stretch shortening cycle performance improvements in endurance runners (8%) compared to untrained controls (12%). As the endurance runners had stiffer tendons than untrained controls, they proposed that endurance runners had an inferior ability to utilise elastic energy, as a result of their stiffer tendons.

Svantesson et al. (136) (Table 3) suggested that increased muscle and tendon stiffness, in especially weaker individuals, had a significant positive impact on stretch shortening cycle potentiation. However, their data indicates that this is not a strong relationship.

## FORCE TRANSMISSION

It may be argued that a stiffer tendon is detrimental to eccentric contraction force and also stretch shortening cycle muscle action. In accordance with this as the eccentric load increases on the muscles, the linear extension of the contractile components would increase (111), thereby reducing the amount of cross-bridge overlap and force generating capabilities of the involved muscle fibres (149;155). Alternatively, Koceja and Kamen (88) speculate that tendon properties may be different amongst individuals and that this difference might effect reflex sensitivity and –contribution to muscle force generation. With a stiffer tendon, it is possible that most of the extension will be transferred to the contractile component of muscle, which increases spindle distortion and afferent feedback (149). This could, in turn, facilitate stretch shortening cycle performance by means of myoelectrical potentiation (149). The eccentric load on the muscle-tendon complex, however, determines the extent of this effect (149), as high eccentric loading induces a greater stimulation of the Golgi-tendon organ (17;104;149) and subsequently attenuates concentric performance (60;149). Therefore it is possible that a stiffer muscle-tendon complex might be beneficial for stretch shortening cycle performance during lower eccentric loading conditions.

Force generated by the muscles, during the initial stages of muscular contraction, requires stretching of the series elastic components (91). Next, the force is transferred to the skeletal system by means of the muscles' tendinous origins and insertions to create movement (109;124). A stiffer muscle-tendon complex, due to its decreased extensibility and decreased time delay due to stretching (27), therefore transfers these forces more rapidly to the bones, contracts more powerfully (155), and facilitates the initial rate of force development (155;156). Tendon stiffness was positively related to improved concentric and isometric muscle performance (Table 3) (155), and stiffer subjects therefore generate more power during pure concentric movements (156). Wilson et al. (156) proposed that a more compliant system allows a faster muscle fibre shortening velocity, thereby limiting its force-generating capabilities. As muscle fibre shortening velocity is slower, and the muscle fibres are comparatively lengthened throughout the active range of motion, a stiffer muscle-tendon complex would be able to generate more force during pure concentric movements (155). Wilson et al. (155) however point out that these advantages of a stiffer muscle-tendon complex decrease as the magnitude of contraction becomes greater for a



stiffer, compared to a more compliant system. They also propose that a stiffer tendon can only be optimal for improved stretch shortening cycle muscle performance if the increased stiffness facilitates the concentric contraction phase by improving reflex activation and force-transmission, more than the loss in stretch shortening cycle potentiation, as a result of the reduction in the contribution from elastic recoil (155).

### ***Summary***

The difficulties in measuring the elastic properties of the muscle-tendon complex has resulted in an unclear understanding of the contribution of this complex to elastic energy restitution and potentiation of muscle performance (133). Although several studies suggest an association between a more compliant muscle-tendon complex, and enhanced stretch shortening cycle performance (Table 3), this interpretation is not conclusive and will need to be tested in further studies.

## **SUMMARY OF THE LITERATURE**

It has repeatedly been shown that by eccentrically pre-stretching active muscle, the ensuing concentric contraction is significantly enhanced. Numerous mechanisms have been proposed to contribute to the potentiation or improvement of concentric performance found during stretch shortening cycle movements:

- The storage and release of elastic strain energy via the stretching and elastic recoil of the tendons and aponeuroses
- Relatively increased muscle activation at the onset of concentric contraction in stretch shortening cycle muscle actions allowing for greater force production
- Higher force output at the onset of concentric contraction in eccentrically pre-stretched muscle leading to increased acceleration
- Myoelectric potentiation of the involved muscle fibres through neural pathways
- Improved force-velocity and length-tension relations of the involved muscle fibres due to the functional interaction between muscle and tendon

To isolate any one of these components as the main contributor to the potentiation would be inappropriate, as the interaction of these mechanisms seems to be instrumental in improving concentric muscle function in normal human locomotion, i.e. during the stretch

shortening cycle. On the basis of this literature review, the following assumptions have been made with regards to stretch shortening cycle enhancement of performance:

- During eccentric stretching of muscle, the reflex pathways are stimulated, which as a protective mechanism, increases neural activation and the formation of additional actin-myosin cross-bridges.
- This facilitates an increase in muscle stiffness and resistance to stretch in the involved muscle fibres.
- This increased muscle stiffness and resistance to stretch boosts the force generated in the muscles, and imposes the majority of the stretch on their respective tendons and aponeuroses.
- Concentric contraction is initiated at a higher force and activation level, which accelerates the body or body part faster and more forcefully, than it would during isolated concentric contraction.
- As a result of the substantial decrease in force from eccentric to concentric contractions, combined with the increased acceleration during the initial part of concentric contraction, the recoil of the previously stretched tendons and aponeuroses, additionally assists in the shortening of the muscle-tendon complex.
- Furthermore, because of the complex interaction between muscle and tendon, the muscle fibres, due to more optimal length-tension and force-velocity relations, also contract more forcibly throughout the movement.
- The combination of events rather than the isolation of any of these abovementioned mechanisms contribute to enhance the power output and mechanical efficiency found in stretch shortening cycle movements.

The extent of interaction or proportional contributions of these events is presently unknown, and it might in fact be pointless to even try and quantify their relative contributions, as they all influence each other in some way and contribute to performance enhancement. Yet, whichever way they may interact with each other, there seems to be overwhelming evidence that tendon elasticity has a profound effect on stretch shortening cycle performance.

Of all the abovementioned parameters tendon elasticity seems to:

- Have a dominant effect in determining the potentiation of muscle performance during the stretch shortening cycle.
- Have a significant effect on myofibrillar displacement during eccentric stretching, affecting both neural reflexes and the length-tension and force-velocity relations of the involved muscle fibres.
- Make an important contribution to the storage and utilisation of elastic energy and could also effect force-transmission and excitation-contraction coupling-time.

To gain a better understanding of muscle function during exercise it is therefore important to gain insight into the elastic characteristics of tendons and tendinous structures *in vivo*. Individual variation in the elastic properties of these structures might additionally broaden the understanding of the mechanisms behind human movement.

## RESEARCH QUESTION

The primary goal of this dissertation is to identify the relationship between the mechanical characteristics of the muscle-tendon complex, in particular tendon stiffness, and stretch shortening cycle muscle function. To fulfil this goal, the main aims of this thesis are:

- (i) To design and develop a testing procedure/equipment that can measure stretch shortening cycle muscle function and musculotendinous stiffness in both upper and lower body musculature
- (ii) To determine if the procedures for measuring musculotendinous stiffness in both upper and lower body musculature are repeatable.
- (iii) To determine if the procedures for measuring stretch shortening cycle muscle function in both upper and lower body musculature are repeatable.
- (iv) To refine the testing procedures, based on the experience gained in (ii) and (iii).
- (v) To determine the relationship between musculotendinous stiffness and stretch shortening cycle muscle function in both upper and lower body musculature.

## **CHAPTER 2:**

### **The building of the equipment - A stepwise description**

#### **INTRODUCTION**

Previous research in the MRC/UCT Research Unit for Exercise Science and Sports Medicine has identified a need for a procedure to measure the stretch shortening cycle of muscle function. This incentive led to an extensive review of the literature on the stretch shortening cycle of muscle function and relative testing procedures.

During the literature review the concept of musculotendinous stiffness and the potential influences this may have on stretch shortening cycle performance was identified. Non-invasive testing for musculotendinous stiffness was described as being performed through either force-oscillation vibration physics (32;80;112;132;149;151-153;155;156), isokinetic testing (136) or ultrasonography (96-102).

The isokinetic testing and isolated ultrasound techniques have limitations from an applied exercise physiology perspective. Therefore, it was decided to design a piece of equipment that could be used to test for musculotendinous stiffness, using the oscillation technique, and stretch shortening cycle performance in a similar manner to that used in previous studies (149;151;153;155;156).

To fulfil other research needs, the equipment needed to be a multi-purpose design and be able to test both the upper and lower body for both musculotendinous stiffness and stretch shortening cycle performance. Also, the equipment had to consider specific nuances. For example, a requirement during testing stretch shortening cycle performance is to adequately control movement so that subjects when jumping with, or throwing the bar, start and stop the movement in approximately the same position for a precise measurement. Also it was important that movement was controlled in one plane. These factors were therefore considered in the preliminary design of the equipment.

It was decided that the force plate and other electronic components be built instead of purchased, as their function needed to be customised. With this in mind, one of the Gym equipment manufacturers in Cape Town, namely Paul Beugelink of Zest Manufacturing PTY (Ltd), was approached to enquire if his company would be interested in a partnership with developmental research concerning new testing and training equipment.

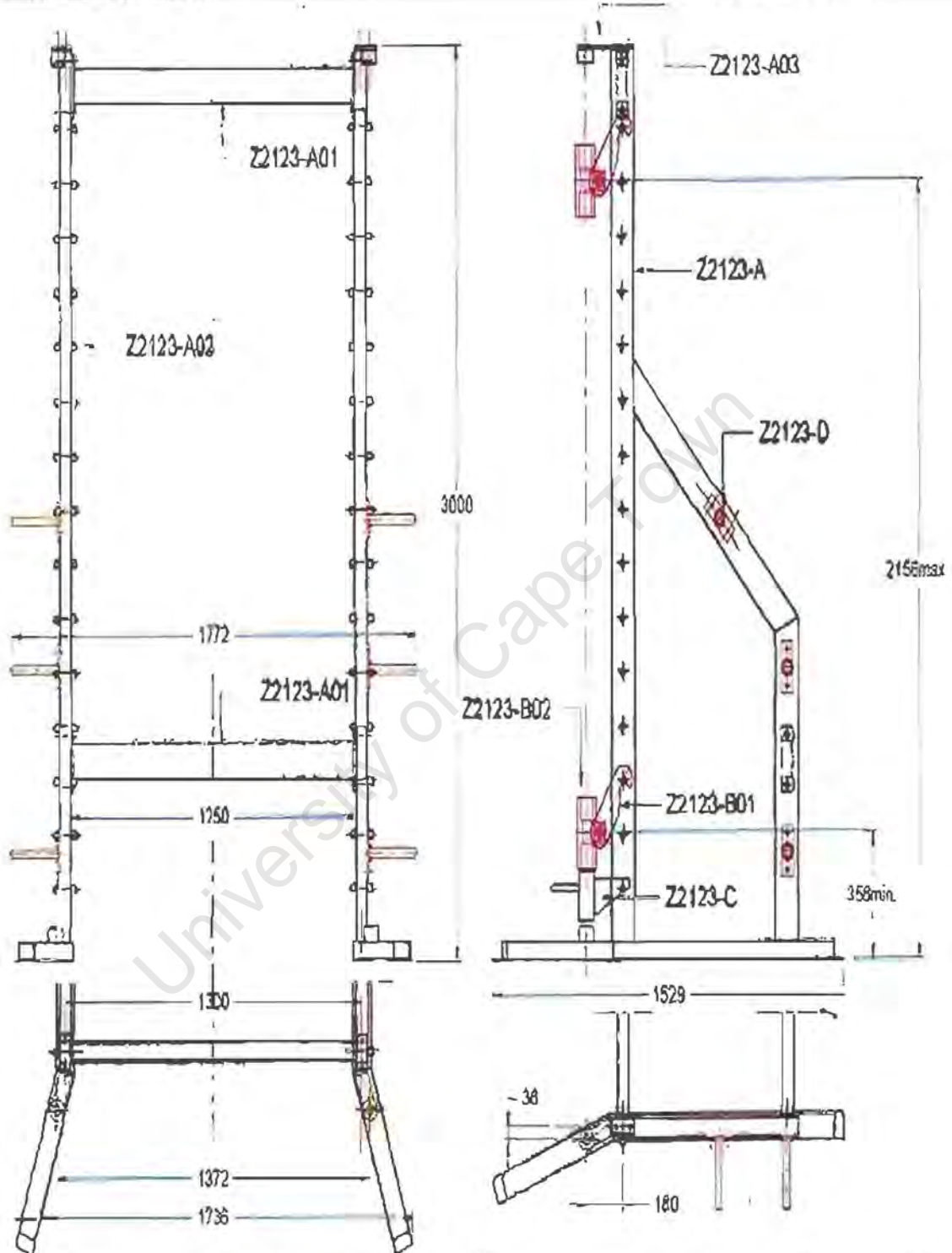
After sharing the idea and regular correspondence, a proposal for the project was developed for Zest Manufacturing PTY (Ltd). They accepted the proposal and agreed to sponsor the construction of the structural equipment, i.e. the materials, the costs, and the labour involved. In return, a working agreement was compiled for both the institutions (MRC/UCT Research Unit for Exercise Science and Sports Medicine and Zest Manufacturing).

To make the equipment a practical and versatile investment, other design features were incorporated to make it available to other research areas for future projects. Although based on an idea from previous research {3}{10}{11}{12}{20}, the equipment was a unique design and has its own features and technology. The following sections are dedicated to the process and development of the equipment.

## **EQUIPMENT DESIGN**

The equipment structure was revised and adapted many times during the initial eighteen months to make certain components more functional and to increase the application ability of the equipment. It is impractical to document all the alterations to the equipment from the beginning to the end of construction, so only the relevant alterations leading to the final draft will be described.

# THE MODIFIED SMITH MACHINE




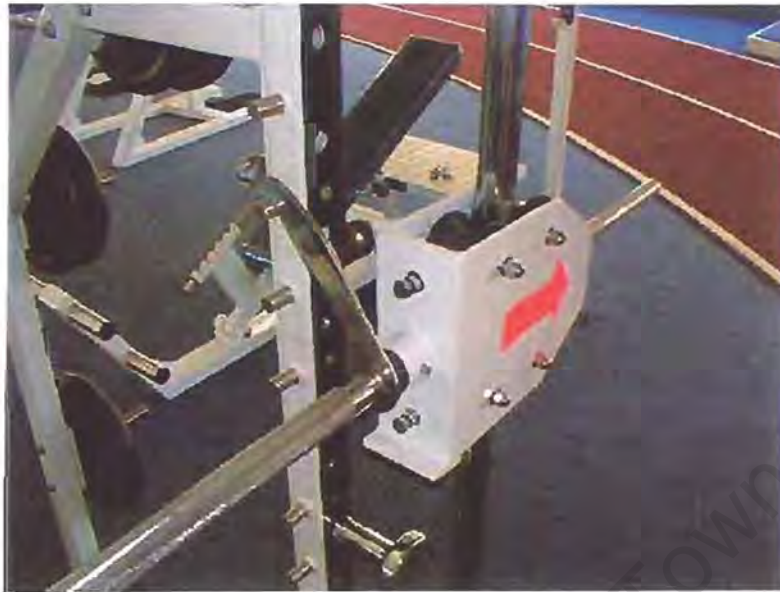
ZEST	copyright notices on all images and remain to property of ZEST Manufacturing (Pty) Ltd. And shall not be disclosed to third parties unless by prior written consent.		scale: not date: 28 June 01	1:1	SMITH MACHINE frame assembly	revision	C	drawing no Z2123 (1)
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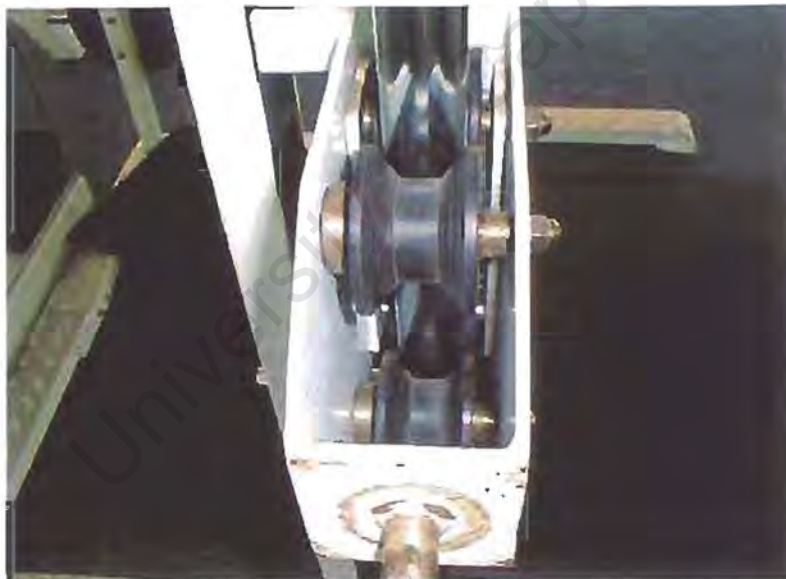
Figure 1: The design sketch for the modified Smith machine.



The original design of the Smith machine, as made by Zest Manufacturing PTY (Ltd), Retreat, Cape Town, South Africa, used an adapted pulley-system to have linear tracking of the horizontal bar on the vertical shafts (Figures 2 and 3).



**Figure 2:** Linear tracking unit of the original design Smith machine, as previously made by Zest manufacturing PTY(Ltd.)



**Figure 3:** Pulley-system of the linear tracking unit of the original Smith machine design by Zest manufacturing PTY(Ltd.)

The whole bar-collar unit originally weighed about 35 - 40 kilograms. For power testing purposes, this design had to be changed to make the bar-tracking unit lighter, as well as strong enough to be able to withstand heavy loads.

Another factor that had to be taken into consideration in the design of the equipment was the resistance created by the pulley system. The linear tracking unit had to be able to

travel vertically up and down the chromed steel shafts with no- or minimal friction. To achieve this the linear tracking unit needed to move as close to friction-free as possible and with no counterbalances (used by many manufacturers) so that neither resistance nor assistance could be given to the vertical forces and bar movement.

In the first prototype of the original linear tracking units attached to the Olympic bar, we used a Vesconite linear tracking unit. This was initially used with the intention of reducing the mass of the system. Vesconite is a self-lubricating polymer that would have reduced the mass of the linear tracking unit substantially and maintained a smooth surface with minimal friction. Unfortunately, this did not work out as it created too much resistance, and in the final design, linear-bearings in a chromed steel housing were used (Figures 4 and 5).



**Figure 4:** *The new design linear-bearing tracking unit with steel housing in the early phases of production.*



**Figure 5:** *The final chromed version of the linear bearing tracking-unit of the modified Smith machine.*

When testing the system, it was further found that the normal rubber seals within the casing of the tracking unit created a vacuum effect. Following an explosive movement



when the bar was released, as well as during the falling phase when the bar returned to the individual catching it, the rubber seals kept on decelerating the bar. After removing the rubber seals, the movement of the Olympic bar and linear tracking units were true and the unit moved up and down the shaft with minimal friction.



**Figure 6:** The modified Smith machine standing 3 meters in height during the early phases of production.

The bar - tracking unit was designed for functions other than just the oscillation technique and ballistic testing. Therefore the design could not sacrifice too much of the integrity of the bar unit for reduction in mass of the system. The Olympic bar was made strong enough to hold very heavy loads for maximal static and dynamic strength testing while being light enough to perform ballistic testing. The effective end-mass of the bar-linear-tracking unit combination was 21.32 kg. This mass was confirmed on the force plate (to be described further on). During the trial to confirm the mass, two subjects were weighed without the bar in a stationary standing position. Each subject then placed the bar on the

shoulders and maintained the load in a still standing position. The mass measured without the bar unit on the shoulders was subtracted from that of the bar-subject mass. The two readings confirmed exactly the same result.

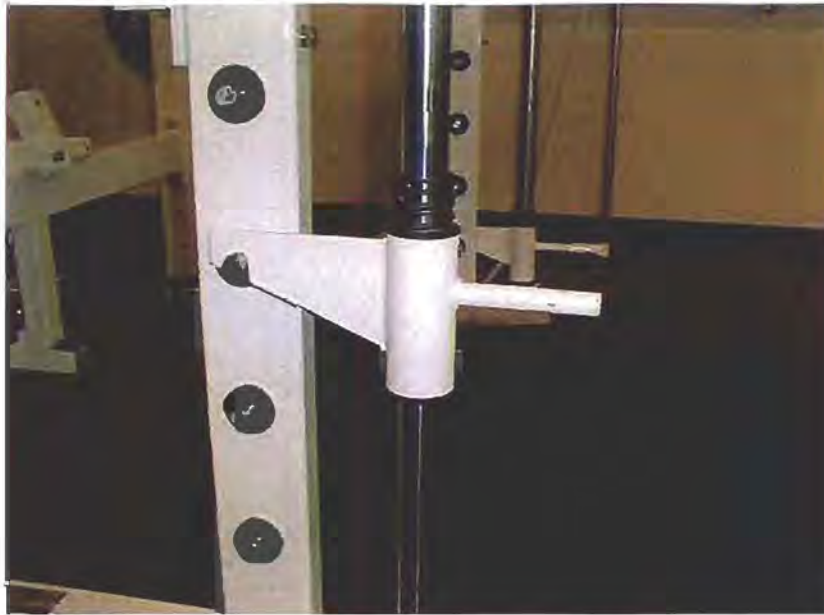
The Smith machine (Figures 1 and 6) was modified to stand three meters high so that proper ballistic testing could be performed. Chemical bolts to limit movement of the system and to further reduce the friction of the bearings on the shafts bolted the structure into the ground. This design catered for testing and training squat jumps, push-presses, plyometric bench presses and many other ballistic exercises.

Other modifications were made for the purpose of both testing and training techniques. For example, top stoppers were added to the Smith machine (Figure 7). These were added for the purpose of isometric and for functional isometric (52;122) testing and training. Lower position stops were added as a standard protective mechanism (Figure 8). If the subject should fail at any time during execution of a test or training repetition, the stoppers would prevent the bar unit from injuring the subject. Double-coiled metal springs were additionally placed on top of the stoppers as extra precaution, should the subject fail at a heavy load.



Figure 7: One of the upper or top stoppers for isometric or functional isometric testing/training.





**Figure 8:** One of the lower stoppers with the double-coiled spring as a protective mechanism for eccentric failure.

Furthermore, an additional sound trigger mechanism (Figures 9 and 10) was attached on both an upper and lower position. This mechanism could be adjusted according to the individual and the type of measurement involved. The lower position is mainly used to standardise the bottom position of any jumps, presses or throws so that one can compare individuals and also compare concentric with stretch shortening cycle effects using the same range of motion and thus depth of movement. This can be applied in both upper and lower body tests and in training.



**Figure 9:** The lower sound trigger used for indicating reversal of direction and standardising the range of movement during testing.

The upper sound trigger was designed for use as a minimum height indicator that the subject had to achieve in a training session. It can also be used as a minimum height indicator for fatigue studies, where the individual has to achieve a minimum height for each repetition to standardise the comparative results.



**Figure 10:** The sound trigger is connected via stretch-cord to a very loud electronic bell.

An additional extension at the base of the back support beam of the modified Smith machine allows the leg press seat section of the equipment to be attached for maximal isometric, functional leg press and oscillation tests by means of a pull-pin lock system (Figure 11). On the topside of the back support beam, another extension (Figure 12) allows the leg press unit to be locked down for maximal isometric tests and for the shaft of the extra arm attachment (Figure 29) to slide through during the functional leg press.



**Figure 11:** The leg press seat-section locks onto the modified Smith machine via a long shaft and pull-pin lock mechanism.





**Figure 12:** *The topside extension of the back support-beam, where the leg press unit is locked onto the modified Smith machine for isometric leg press tests.*

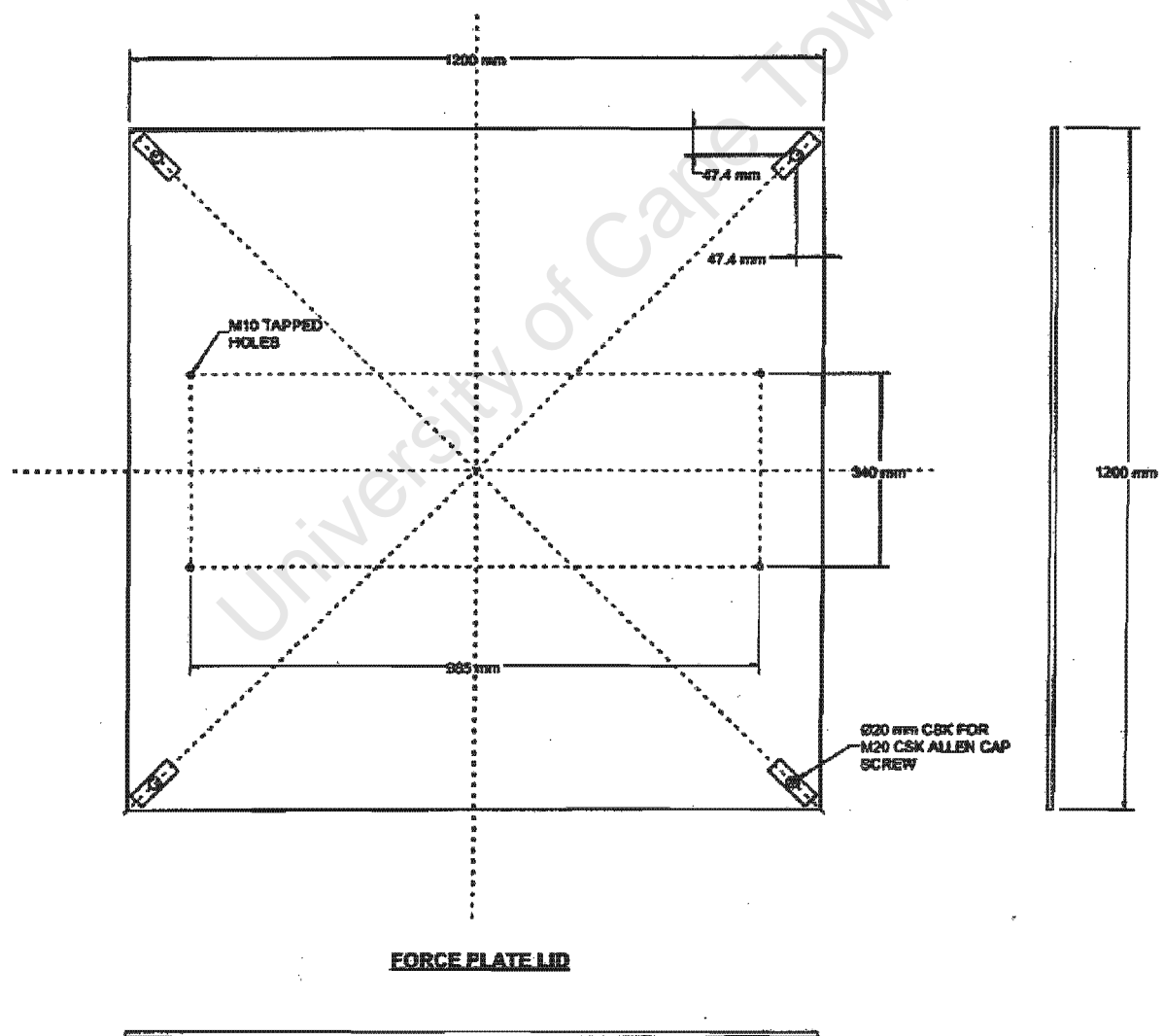
At the back of each of the upright supports of the modified Smith machine, a further extension allows the leg press unit to be attached, also by means of a pull-pin lock system.



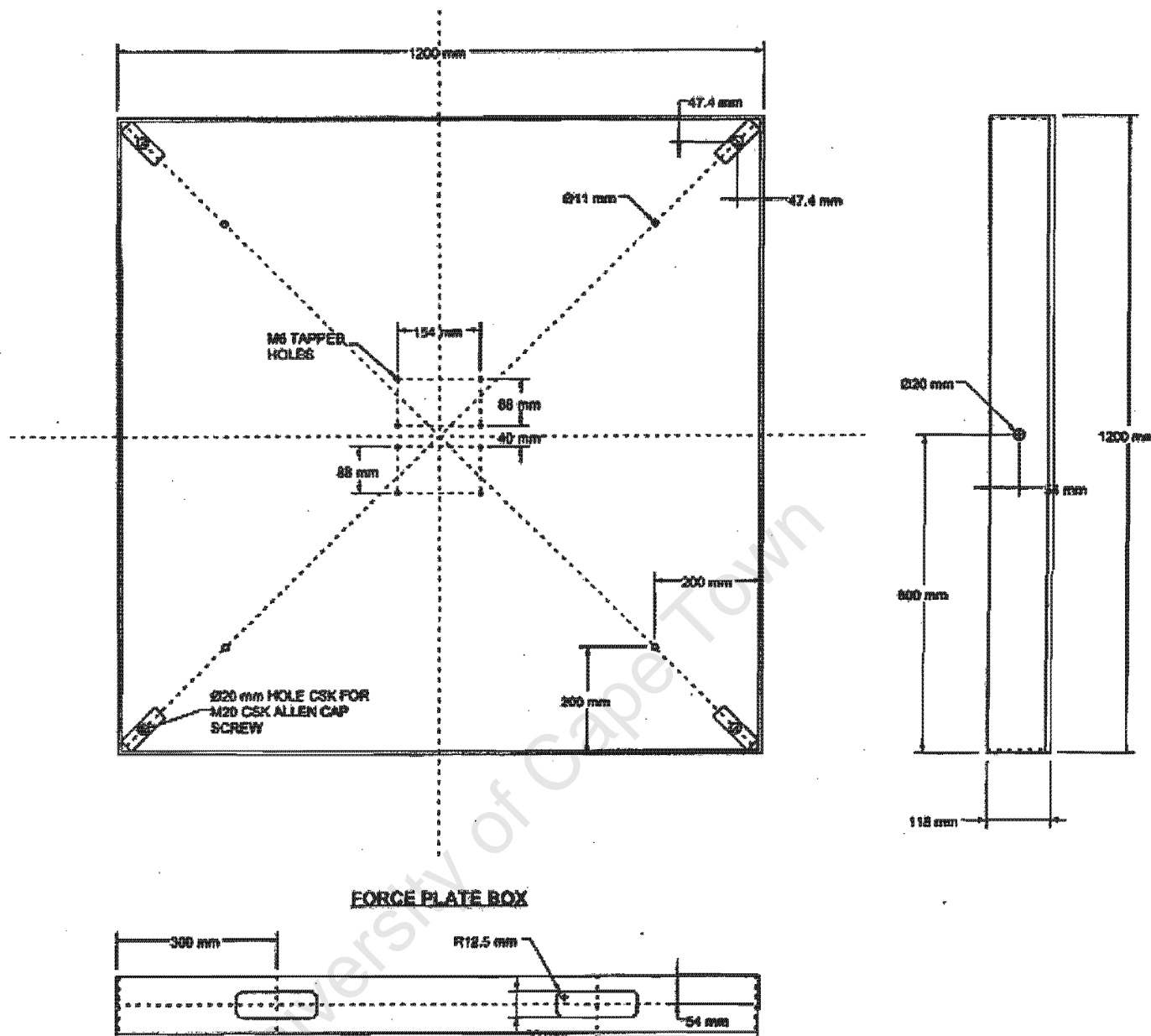
**Figure 13:** *The extension attachment at the back of the upright supports of the modified Smith machine where the leg press unit is mounted.*

## FORCE PLATE

The cost of a force plate with the dimensions needed for the study exceeded the budget. Hence, the task to design, customise and build a force plate was the next phase in the making of the equipment. While the main structure was being assembled, research into force plates and their sub-components was underway. Using the gait analysis force plate in our laboratory as a basis and researching details on various force plate technology providers such as Kistler, AMTI and Bertec, we ascertained what was needed to build our own functional force plate with the capabilities of measuring force in the range of 0 N to 78.48 kN with a high accuracy and sensitivity of measure (refer to Figures 14-24).

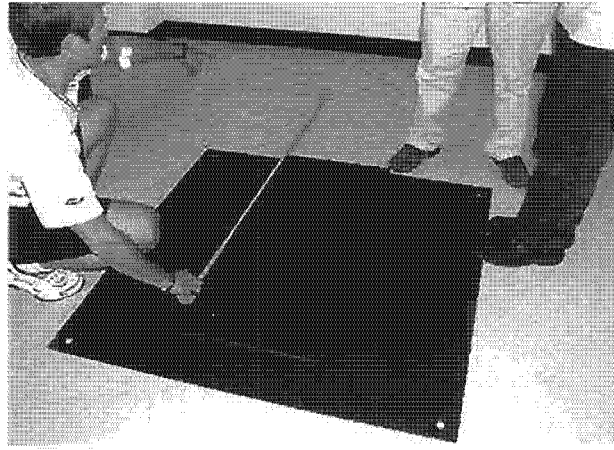


**Figure 14:** The design sketches of the force plate lid.



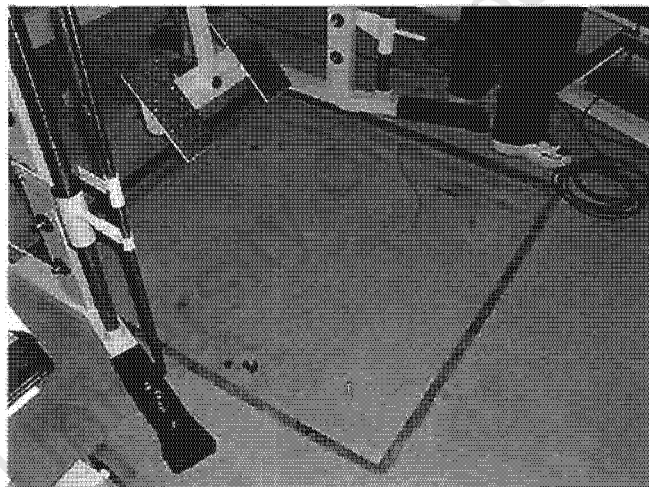
**Figure 15:** The design sketches of the force plate box without electronic components.

Initially there was a need to determine where the equipment was going to be based. After consulting the Sport Science Institute of South Africa's structural engineers, Liebenberg and Stander Consulting Engineers, a site was chosen for the equipment. They recommended that the original screed layer on the floor surface be removed and a concrete slab be laid down with the force plate sunk into it, rather than sink the force plate directly into the floor. This process was then started and the concrete slab was constructed using the dimensions of the main structure and its sub-components and taking into account the size of the force plate (Figure 16).



**Figure 16:** Before the concrete slab was constructed, measurements of the dimensions of the force plate, movement space, the placing of the NAMS (Neuromuscular And Musculotendinous Stiffness) Unit and were calculated.

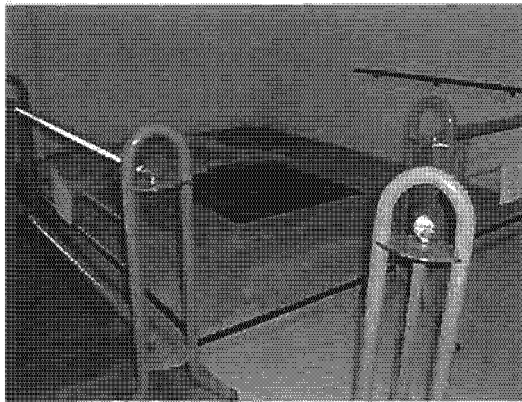
On the bottom concrete layer, the slab was reinforced with wire mesh to make the slab strong enough to tolerate high loading and impact forces. On the top layer, self-levelling screed was used to make the top surface flat and exact.



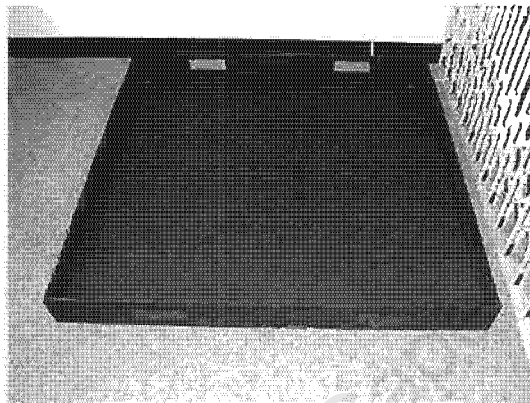
**Figure 17:** The hole into which the force plate was placed. Also visible is the beginning of the electrical conduit where all the wiring went through to the power supply.

A hole was made in the middle of the slab according to the dimensions of the force plate, so that the top surface of the force plate would be flush with the top surface of the slab (Figures 17 and 18). An electrical conduit was placed through the slab, leading from the force plate to the wall, and further along the wall until it exited the slab near the electrical power supply. After the slab had dried sufficiently and had reached maximum strength, the main equipment was bolted onto it and the slab was further covered in carpeting.



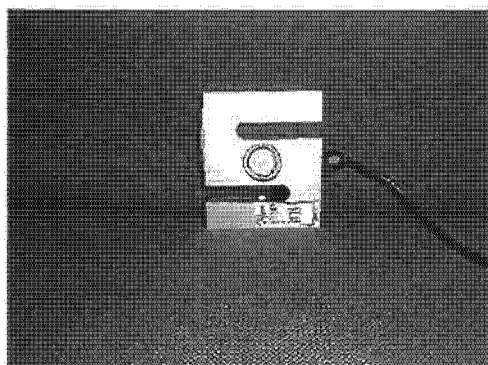


**Figure 18:** The final, fully dried concrete slab with the force plate box in the hole before the equipment and carpeting was added.



**Figure 19:** The force plate box base without the top plate before the electronic components and load cells were added.

After consultation with various biomedical, electrical and mechanical engineers at the University of Cape Town, we contacted Dr Danie van Vuuren at Route Industrial Automation, Johannesburg, South Africa. We asked him to make us five RSS-type load cells, each with a two tonne maximum load capacity and a high sensitivity range (Figure 20). The reasoning for the high load capacity of the load cells was so that the system could also be used for testing other measurements such as maximal strength and power for all ranges of athletes.



**Figure 20:** An RSS-type 2 tonne capacity load cell with its cable feed, which connects to the junction/summation box and strain gauge amplifier

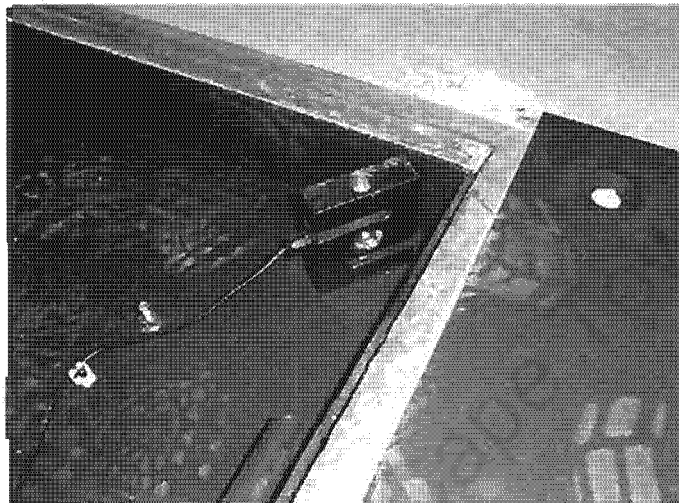
Each load cell has the following specifications:

Specifications	Values
<i>Rated capacity (RC)</i>	2000 kg
<i>Rated Output (R.O.)</i>	2 mV/V = 0.5%
<i>Non-linearity</i>	0.03% R.O.
<i>Hysteresis</i>	0.03% R.O.
<i>Repeatability</i>	0.02% R.O.
<i>Zero balance</i>	± 1% R.O.
<i>Temperature range, compensated</i>	-10-50°C
<i>Temperature effect on rate output</i>	± 0.1% LOAD/10°C
<i>Temperature effect on zero balance</i>	± 0.05% R.O./10°C
<i>Terminal resistance, input output</i>	350* ± 3.5* 350* ± 5*
<i>Insulation resistance (Mlr.)</i>	
<i>Bridge to ground</i>	2000 M*
<i>Shield to ground</i>	1000 M*
<i>Excitation recommended</i>	10V
<i>Excitation max</i>	15V
<i>Safe overload</i>	200% RC
<i>Height of load cell</i>	108 mm
<i>Thickness of load cell</i>	28.6 mm
<i>Depth of load cell</i>	88.9 mm
<i>Tapped hole dimensions</i>	M20 x 1.75

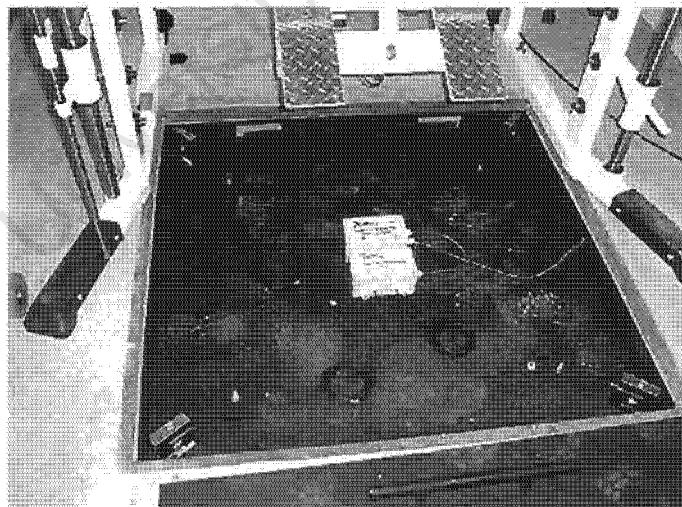
One of these load cells were incorporated into each of the four corners of the force plate and a fifth load cell was used as a stand-alone system in the leg press unit.

The force plate metal box and M20 Allen-cap bolts for the load cells were also sponsored and made by Zest Manufacturing PTY(Ltd). Because of the size of the force plate, i.e. 1.2 m x 1.2 m x 0.108 m, the base plate and top plate were both made of solid steel and for increased stiffness were 10 mm thick. This made the force plate very heavy. The base (± 130 kg) was bolted onto the slab by means of chemical bolts to limit its vibration and movement (Figure 21). The load cells were also bolted on the base plate using

countersunk Allen-cap bolts that were further maintained in position by using an anaerobic adhesive called Professional Fix-lock, which was applied to the bolts before tightening (Figure 21). Before the base plate was placed in the hole, the load cells were tested for any movement. It was confirmed that there was total contact between the load cells and the base plate and that no movement between them was allowed. We also had a junction/summation box manufactured that sums the return signal of the four load cells and transmits the summed voltage reading to a pre-amplifier (Figure 22).



**Figure 21:** An RSS-type load cell bolted down on the base plate of the force plate with the top plate visible on the right. Also visible is the chemical bolt that was used to bolt the base plate down.



**Figure 22:** The base plate of the force plate with all four load cells bolted down and connected to the junction/summation box and strain gauge amplifier. The cables from the strain gauge amplifier leave the force plate through the electrical conduit to the power supply and the computer system respectively.

Originally, we had a 6000 Hz high-speed transmitter with a 0 - 10 Volt re-transmission signal and built-in pre-amplifier attached to this system. However, the transmitter was

creating too much high frequency noise and even after filtering, the signal was not very pure. After having an electrical technician inspect the system, it was determined that the transmitter was the origin of all the noise. After bypassing the transmitter and re-routing the signal straight through the pre-amplifier, it was found that although the signal was much clearer, the system was overheating and that the amplifier would not be able to maintain this voltage input for prolonged periods. It was then decided to try and find a standard strain-gauge pre-amplifier that would be able to drive the system without overheating and not have to run the amplified signal through a transmitter. After various searches, a strain gauge amplifier and compatible PC-board was found through RS Components International PTY(Ltd.), Montague Gardens, Cape Town, South Africa. We built up the PC-board and then tested it on the system (Figure 23). It tested positively allowing us to move to the next phase of testing the equipment. This will be explained later on in the chapter.

The technical specifications of the strain gauge amplifiers used, are as follows:

Specifications	Values
<b>Supply voltage</b>	$\pm 2 \text{ V to } 20 \text{ V DC}$
<b>Input offset voltage</b>	$200 \text{ } \mu\text{Vs (max)}$
<b>Input offset voltage drift</b>	$0.5 \text{ } \mu\text{V/}^\circ\text{C(temp),}$ $0.3 \text{ } \mu\text{V/month(time), } 3 \text{ } \mu\text{V/V(supply)}$
<b>Input impedance</b>	$> 5 \text{ M}^*$
<b>Bandwidth</b>	$450 \text{ kHz (unity gain)}$
<b>Output voltage</b>	$\pm V_s - 2 \text{ V}$
<b>Output current</b>	$5 \text{ mA}$
<b>Closed loop gain</b>	$3 \text{ to } 60\,000 \text{ adjustable}$
<b>Open loop gain</b>	$> 120 \text{ dB}$
<b>Common mode rejection</b>	$> 120 \text{ dB}$
<b>Bridge supply temperature</b>	$20 \text{ } \mu\text{V/}^\circ\text{C}$
<b>Bridge supply current</b>	$12 \text{ mA (max)}$
<b>Power dissipation</b>	$0.5 \text{ W}$
<b>Operating temperature</b>	$-25 \text{ to } +85^\circ\text{C}$

Once the base, junction/summation box and strain gauge amplifier were complete, connected and the load cells zeroed, the top plate was bolted directly onto the load cells. Six sets of M10 tapped hole pairs were made on the top plate. These holes were made so



that the bench press bench could be bolted onto the top plate of the force plate at three different settings, depending on the size and morphology of the subject being tested.



**Figure 23:** The two RS Components International strain gauge amplifiers used for the force plate and leg press units mounted on PC-boards.

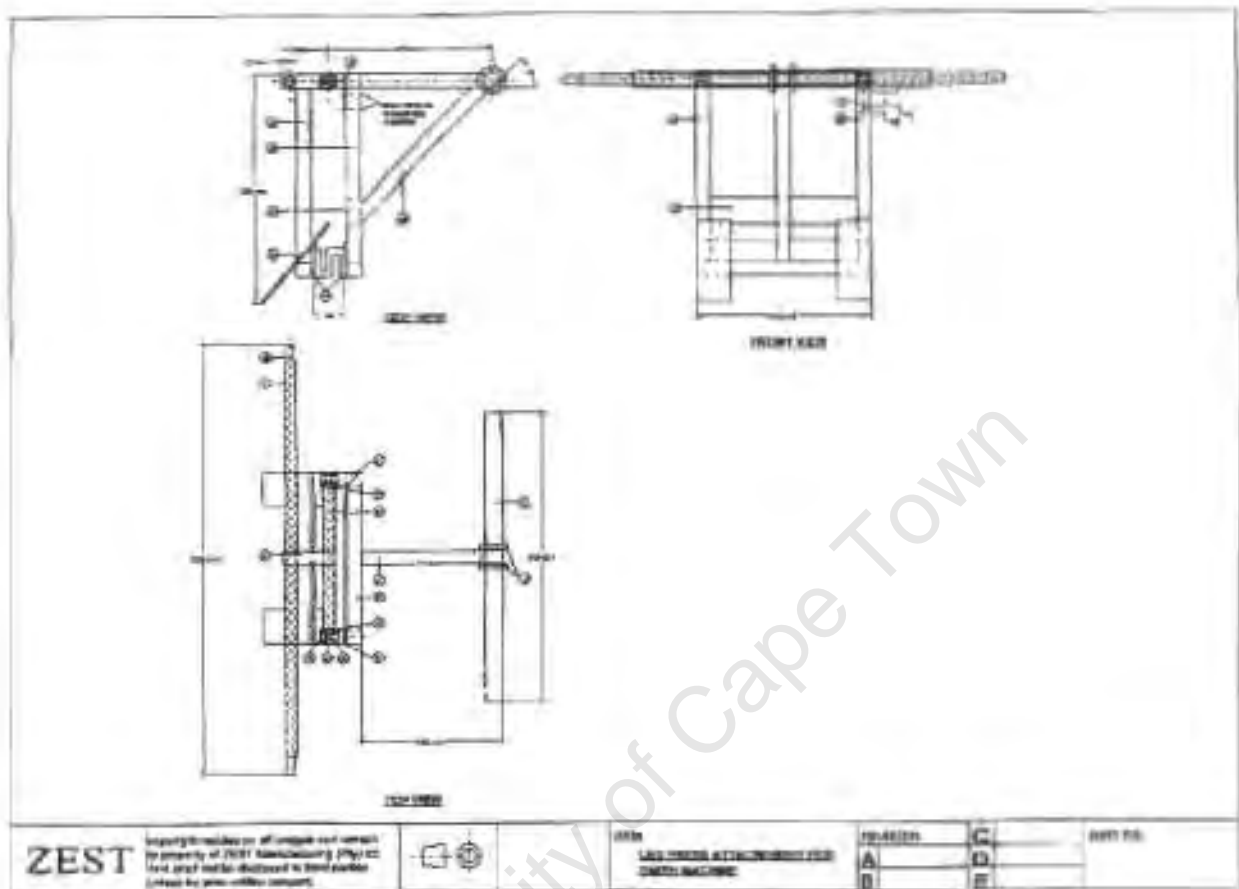


**Figure 24:** The final force plate with the top plate bolted onto the load cells with countersunk Allen-cap bolts, together with the equipment bolted onto the slab. Also visible are the six sets of M10 lapped-hole pairs

## **THE LEG PRESS UNIT**

The original design of the equipment used by Walshe et al. (151), had a leg press unit attached to the Olympic bar by a clamp mechanism. This allowed the unit to rotate with the bar in its sleeve. We decided that this structure needed to be able to function on its

own and be separate from the bar-tracking unit of the modified Smith machine, thereby increasing the durability and versatility of the system



**Figure 25:** The leg press unit, which attaches to the modified Smith machine



**Figure 26:** The leg press unit attached to the modified Smith machine

The leg press unit (Figures 25 and 26) was therefore designed to attach to the modified Smith machine at the previously mentioned extensions at the back of the vertical upright supports with the ability to be adjusted up or down, depending on individual morphology (Figure 26). A solid steel bar with Oilon self-lubricating bushing allowed free rotation of the entire leg press unit along the sagittal plane and connected the two extensions of the vertical supports on the modified Smith machine (Figure 26). This solid bar thus allows the leg press unit to rotate through the centre space of the modified Smith machine.

The leg press unit (Figures 25 and 26) consists of two separate subsections, which are connected by means of another solid steel bar and Oilon bushing system at the pivot point between the two vertical arms of the relative subsections (Figure 27). The only thing keeping the two vertical arms of the relative subsections from pressing together in the sagittal plane is a two tonne capacity RSS type load cell (Route industrial automation, Johannesburg, South Africa), which is bolted onto the system by means of two lock-tight Allen-cap bolts (Figure 28).



**Figure 27:** The pivot point of the vertical arms of the two subsections of the leg press unit.

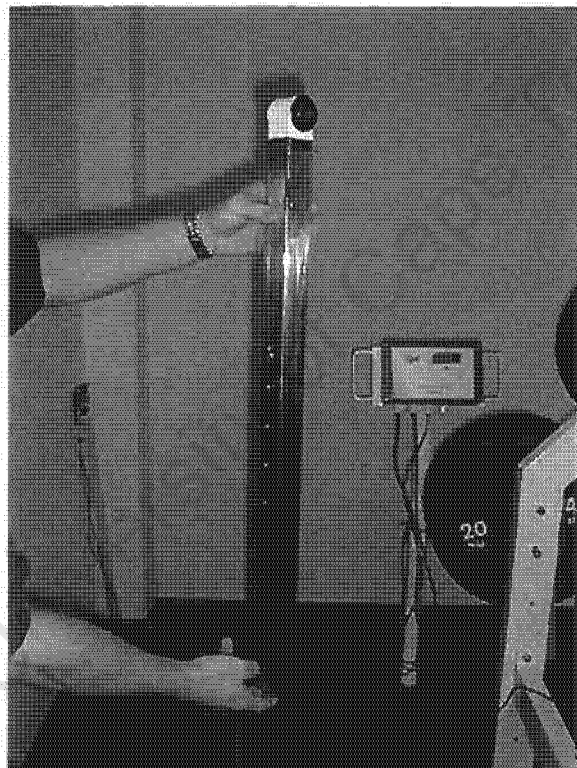


**Figure 28:** The RSS-type load cell between the two vertical arms of the leg press unit measures the compressive force between the two arms



The load cell therefore measures the compressive force between the vertical arms of the two respective subsections. The load cell was originally also connected to a Route amplifier/transmitter, in this case a Route 2500/2700 indicator/transmitter with a 4 - 20 mA and 0 - 10 V re-transmission signal. This transmitter also created excess noise, and burnt out after only a few trials and also having had the raw signal re-routed through the pre-amplifier. Once again an RS Components International PTY (Ltd) strain gauge amplifier and built-up PC board were utilised with great success.

The leg press unit was also designed to measure isometric function. This was achieved by an extra arm attachment, which connects the leg press unit and the Smith machine base by means of two pin-locks (Figures 29 and 30).



**Figure 29:** *The arm attachment that is used to lock the leg press unit to the modified Smith machine back-support beam.*

For functional (e.g. 1RM) isotonic tests, this same arm attachment is used, but only the top pin of the leg press unit is locked. This allows movement of the leg press unit as per normal, but serves to prevent the leg press unit from moving into the subject and injuring him/her during muscle failure. This serves purely as a protective mechanism or stopper.





**Figure 30:** *The leg press unit locked onto the back support-beam of the modified Smith-machine by means of the extra arm attachment.*

During the oscillation technique, the leg press unit moves freely without any stoppers, with the subject carrying the entire load during the test. A leg press seat-section that can be incorporated onto the system was also designed for the leg press unit.

### **LEG PRESS SEAT SECTION**

A reclined leg press seat was designed in a shape that can be adjusted to a position that can simulate a squat position and movement without the same loading on, or risk of injuring, the back. The backrest was also designed to be adjusted up or down depending on the test performed (Figure 31).



**Figure 31:** *The leg press seat-section with adjustable backrest and shoulder pads.*

Shoulder stoppers were also added to stabilise and prevent slippage of the body when loaded. These shoulder pads can also be adjusted according to individual morphological differences. Additional Velcro straps were incorporated to stabilise the torso, support the lower back and isolate the hips and leg musculature (Figure 32).



**Figure 32:** *The Velcro-straps are used to stabilise and support the lower back and hips and isolate the legs and hips for testing*

The seat section can further attach to the modified Smith machine base support-beam as previously mentioned by means of a pull-pin lock system. The leg press seat can be moved either closer or further away from the leg press unit depending on the test performed, individual leg length and hip- and knee angle requirements. The leg press seat-section moves over the top plate of the force plate and connects via a long shaft to the base support beam (Figure 11).

### **BENCH PRESS BENCH**

The bench press bench is standard and conforms to international criteria. It has however an additional feature of being able to function as an incline bench by adjusting the back support upwards, or as a normal flat bench by leaving the back support down on the bottom position (Figure 33). This is however a common feature of numerous bench-press benches.





**Figure 33:** *The adjustable bench press bench used for upper body testing.*

Specific bolts were made so that the bench can be bolted down onto the force plate, specifically for upper body testing. These bolts prevented lateral or multidirectional movement/influence, and reduced the noise created by the bench vibrating and moving on the force plate top plate (Figure 34).



**Figure 34:** *The adjustable bench press bench bolted down onto the force plate to limit movement artefact and increase accuracy of measure.*

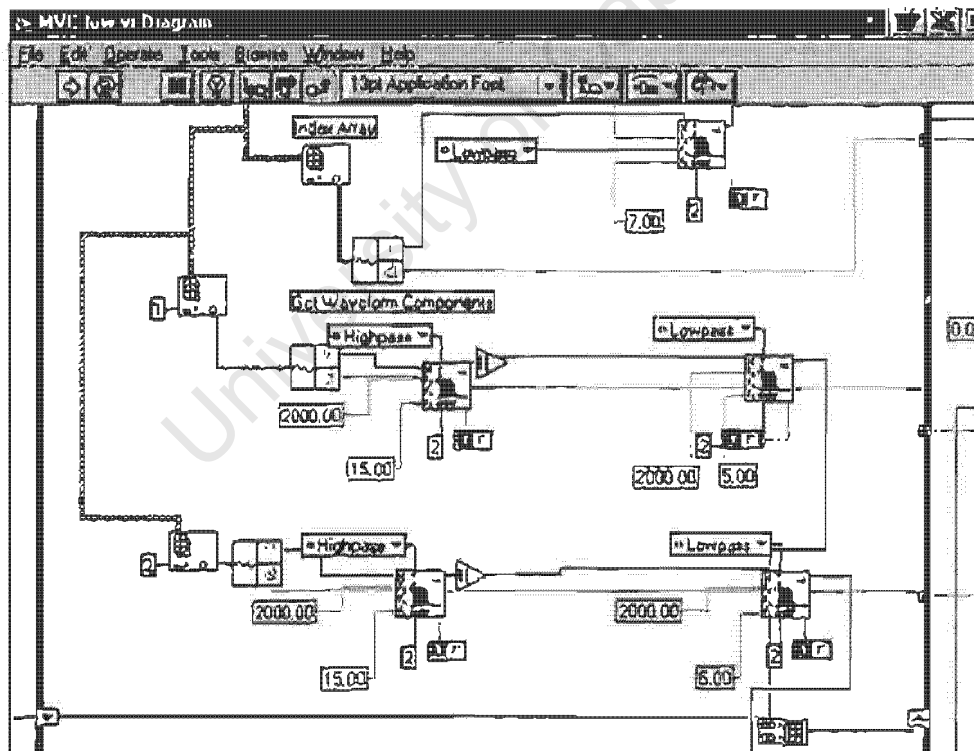
The stiffest and hardest Gym equipment upholstery available were chosen as padding materials for the leg press seat and bench press bench. This material was chosen to

minimise its influence on damping the oscillations during the oscillation tests of the upper and lower body.

## COMPUTER INTEGRATION

The next challenge in developing the equipment for this project was to find software and a computer system that would be able to coordinate the data recording applications. After various computer searches, communications and attending various workshops, demonstrations and seminars it was decided that National Instruments' software would be the best software for this system (LabVIEW Full Development System Version 6.0, National Instruments, Austin, Texas, United States of America).

LabVIEW uses a computer programming language called G-programming. It is a graphical, icon-based language, which is mainly used for engineering applications (Figure 35).



**Figure 35:** An example of the G-programming language used in order to program in LabVIEW 6.0.

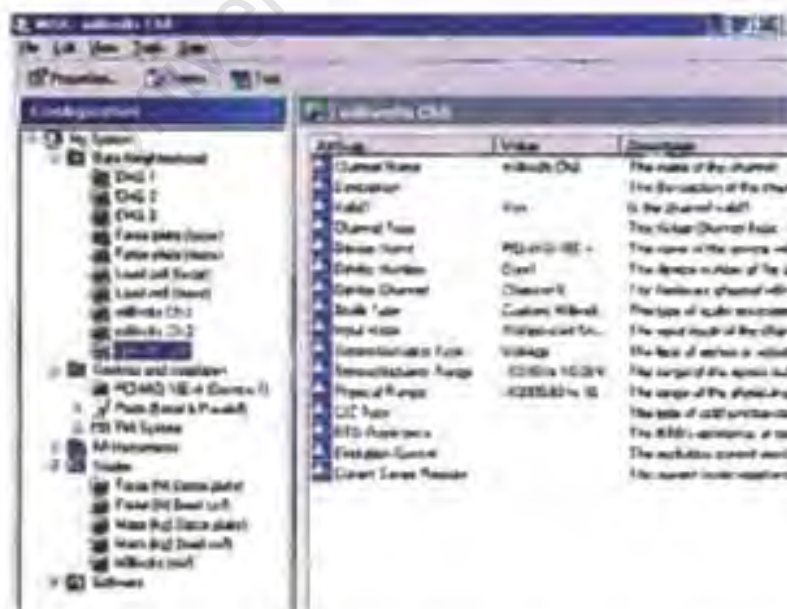
The main two books used to assist in developing the data-acquisition programmes were, "LabVIEW programming, data acquisition and analysis" and the "Hands-on Exercise Manual" by Jeffrey Y Beyon (Prentice-Hall PTR, Upper Saddle River, New Jersey, U.S.A.).

Together with the software we purchased a computer with the specifications received from the National Instruments' website and also ordered an analogue to digital converter card, ribbon cable and connector-block (National Instruments) that was needed to connect and time-integrate the force plate and load cell data (Figure 36).



**Figure 36:** The National Instruments' connector-block mounted to the wall with the cables from the EMG, load cell of the leg press unit and the force plate as they connect to it. Also visible is the computer connector socket, by which the connector block transmits the signals to the A/D card.

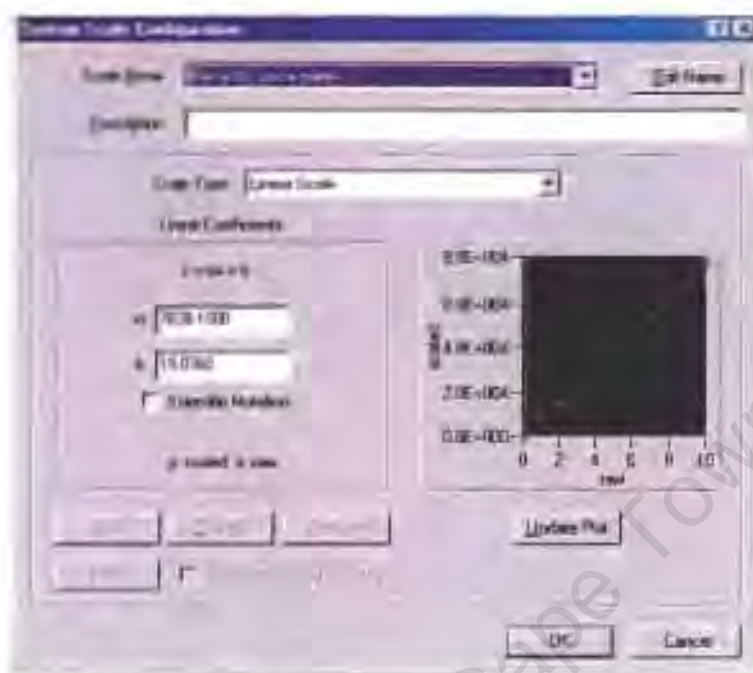
Once the basics of LabVIEW were learnt, numerous tests between the software and the National Instruments' plug-and-play analogue-to-digital converter (A/D) card (PCI-MIO-16E-4) were performed. To convert the raw voltage signal into a standard measure, the system had to be configured by Measurement and Automation Explorer Version 2.0.3.6 (National Instruments software) (Figure 37).



**Figure 37:** The various data channels and scales as set up using Measurement and Automation Explorer. Also visible is an example of the highlighted channel's configuration set-up



The input analogue signal (i.e. the strain on the load cell) is converted to a digital value (voltage) and the voltage signal is then scaled to represent the measure (e.g. force/mass/millivolts) depending on which measurement was being recorded (Figure 38)



**Figure 38:** An example of the scaling factor equation and linear graphical relationship between Voltage and Force output of the force plate

These relationships were determined by connecting an oscilloscope and Voltmeter to the load cell and force plate outputs respectively. The first calibration was performed on the load cell of the leg press unit. With zero load on the load cell the zero point of the load cell was calibrated. Using twelve incremental loads and measuring the frequency noise through the oscilloscope and the voltage through the Voltmeter, a relationship was determined between the voltage and the load. Using a linear regression equation on a statistical software package (Graphpad Instat Version 4.0, Graphpad software Inc., San Diego, CA, U.S.A.), a line of best fit and correlation coefficient were determined to define this relationship. The correlation coefficient ( $r$ ) was calculated as  $r = 1.0000$  and  $p < 0.0001$  indicating a perfect linear relationship between voltage and load. The relationship was calculated as being approximately 5 mV per kilogram (or 5 mV per 9.81 Newton). The relationship was calculated using the equation

$$Y = mX + b$$

Where:

- $Y$  = force
- and  $X$  = voltage

The scaling factor for the leg press unit using the above mentioned equation was calculated as:

$$Y (\text{Force in Newton}) = 1956.2 X (\text{Voltage in Volts}) + 5.531$$

The results of the regression analysis are represented in the following table:

Parameter	Expected value	Standard error	Lower 95% CI	Upper 95% CI
Slope	1956.2	3.372	1948.7	1963.7
Y intercept	5.531	2.399	0.1856	10.878
X intercept	-0.002827			

The mass of the leg press unit was measured on a construction scale as being 45 kg. However, because the unit is pin-locked onto the modified Smith machine and functions as a pendulum, its effective mass was different. Therefore using a scaling factor, the effective mass of the leg press unit was measured by holding the unloaded leg press unit in a stationary position and measuring the compressive mass on the load cell. The leg press unit was calculated as having an unloaded mass of 34.6 kg.

A similar process of calibration was performed on the force plate. The load cells of the force plate were zeroed before the top plate was bolted down and then a similar load-voltage relationship was determined using twelve different loads. A relationship of approximately 1.25 mV per kilogram (or 1.25 mV per 9.81 Newton) was calculated. Using the same statistical software, a linear regression analysis revealed a correlation coefficient of  $r = 0.9999$  and  $P\text{-value} < 0.0001$ , also confirming a near perfect linear relationship.

The scaling factor for the force plate using the above mentioned equation was calculated as:

$$Y (\text{Force in Newton}) = 7638.1 X (\text{Voltage in Volts}) + 19.036$$

The results of the regression analysis are represented in the following table:

Parameter	Expected value	Standard error	Lower 95% CI	Upper 95% CI
Slope	7638.1	24.462	7583.6	7692.6
Y intercept	19.036	6.737	4.026	34.047
X intercept	-0.002492			

Once the force plate and load cell were calibrated and the software scaling factors calculated, the software programming needed completion before the final pilot testing of the equipment and testing procedures could be performed.

For the equipment to function properly, particularly for the measurement of pre-activation, corresponding muscle activation or neuromuscular reflexes, it was necessary to synchronise force data and EMG data. A new telemetric EMG system using the Noraxon Myoresearch 2.02 software, was purchased from Noraxon USA, Inc. However, when installing the Noraxon hardware on the computer system, there was a hardware conflict with the National Instruments' software. Therefore alternative arrangements were necessary. The Noraxon software was also supposed to have LabVIEW support which could be linked to LabVIEW version 5.0 software. Unfortunately we found that the LabVIEW support files with version 6.0 software corrupted the National Instruments' virtual instrument files and therefore created a problem that needed alternative arrangements.

As we already had a National Instruments A/D card and connector block, together with LabVIEW 6.0 software, we connected the Noraxon EMG hardware directly to the National Instruments' hardware and software. Using the Noraxon Telemetry receiver, transmitter, antenna, power supply and electrode cables, we decided however, rather to build an A/D interface cable and interface it directly from the EMG receiver into the National Instruments' connector block. The signal would therefore be transmitted through the National Instruments' PCI-MIO-16E-4 A/D card and avoid using the Noraxon computer hardware and software. After a few trial runs on the LabVIEW software, we found that this worked well.

After successfully programming LabVIEW, the next task was to be able to perform signal processing. Up until then, we could only receive raw force data and raw EMG data. The next step in the programming was to program filters in for the signal processing and to write the data to a spreadsheet file for further analysis.

For the force data, a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz was designed and programmed into the software. For the EMG however, the process of getting a linear envelope for EMG integration analysis was more complex. After many programming changes and trials, we eventually managed to record a linear



envelope after filtering the raw EMG signal. The raw signal was first processed through a second order highpass Butterworth filter with a cut-off frequency of 15 Hz (86), then full-wave rectified (86) by creating an absolute value of each data-point, and finally processed through a second order lowpass Butterworth filter with a cut-off frequency of 5 Hz (86). This finally gave the linear envelope for EMG integration. The equipment, computer hardware and software were finally ready for pilot testing.

## **STANDARDISATION OF TESTING PROCEDURES**

The next phase of the process of getting the equipment validated was to determine the criteria and positioning set-up for the various tests that could be performed on the "Neuromuscular And Musculotendinous Stiffness" (NAMS) Unit. The various tests that could be performed on this new equipment, which are applicable to the present proposed studies were as follows:

- Functional 1RM (one repetition maximum) leg press
- Functional 1RM (one repetition maximum) rebound bench press
- The oscillation test of the upper body
- The oscillation test of the lower body
- MVC (maximal voluntary contraction) of the lower body i.e. leg press
- MVC (maximal voluntary contraction) of the upper body i.e. bench press
- Isolated concentric squat jumps
- Countermovement jumps
- Pure concentric bench throws
- Rebound bench throws

Numerous other strength and power tests could be performed on this equipment, but they are not relevant to this thesis and are therefore not mentioned.

### **THE FUNCTIONAL 1RM LEG PRESS TEST**

Because there was no documented precedent or similar piece of equipment to compare with, we tried to set up the test in a starting position similar to the seated leg press exercise. After repeated trials at different positions and taking the safety of the individual and dimensions of the equipment into consideration, the angles of  $\pm 90^\circ$  at the knee and

$\pm 95^\circ$  in the hip were chosen as the starting position of the leg press. The incline of the backrest was set at  $140^\circ$  or at a  $40^\circ$  incline. A goniometer was used to control the joint angles (Figure 39).



**Figure 39:** A goniometer that is used for measurement of joint angles.

The arm attachment (Figure 29) was pin-locked at the top setting of the leg press unit (Figure 30) thereby using the shaft as a protective stopper mechanism against muscle failure. For the 1RM test, the subject was instructed to press the load up until the legs were straight and then to control the load back to the starting position. During this procedure, the subject was strapped down at the hips using the Velcro-straps (Figure 32) to isolate the legs and protect and support the back. Arms were folded across the chest to remove additional assistance from the arms and back in the movement.

### **THE FUNCTIONAL 1RM REBOUND BENCH PRESS TEST**

For this test the bench press bench was bolted onto the force plate (Figure 34). The subject lay flat on the back with the feet lifted and resting on the edge of the bench. The subject was set up so that the bar of the modified Smith machine moved in the line of the nipples or  $\pm 2 - 3$  cm superior to the xyphoid process of the sternum. The subject was then stabilised on the bench by strapping the lower back down onto the bench press bench using one of the Velcro-straps (shown in Figure 32). The other Velcro-strap was used as a limiting strap around the upper chest area to prevent the lifting of the chest. This strap was however not strapped on too tightly and therefore does not limit chest expansion during maximal inspiration.

The subject's grip positioning was standardised as the width of the arms 90° abducted and elbows 90° flexed i.e. the distance from elbow to elbow. The lower sound trigger (Figure 9) was set to standardise the bottom position of the 1RM rebound bench press test. The sound trigger was set so that it signalled reversal of direction when the bar was  $\pm 6$  cm from the chest. This gave the subject enough time to react and reverse the direction of push within  $\pm 2 - 3$  cm. The subject had to quickly lower the bar until the sound trigger was heard. He/she then had to immediately reverse the direction of movement and push the weight upwards until the arms were locked directly above the chest. Using the sound trigger as a control mechanism standardised the movements and increased the potential for repeatable testing.

As protection against injury, the lower stoppers (Figure 8) were also set up so that the bar never made contact with the chest during the eccentric phase of the movement. If the subject could not lift the weight any further due to muscle failure, he/she was protected against injury, because the weight of the bar was supported and/or absorbed by the stoppers and double-coiled springs. If during any attempt, the subject did not reverse the direction of movement in time and the bar made contact with the stoppers or springs, the trial was discarded and had to be repeated.

### ***THE OSCILLATION TEST OF THE UPPER BODY EXTREMITIES***

During the pilot testing, all formulae were verified and refined by Mr. Trevor Cloete (Engineering Department, UCT), a vibration physics specialist.

For this procedure to work properly, the bench press bench was bolted onto the force plate and the subject set up, as previously described for the 1RM rebound bench press test. In the literature (153;155;156) this test has previously been performed using certain percentages of 1RM rebound bench press, i.e. 15, 30, 45, 60 and 70%. However, because of the load of the bar-tracking unit having a total mass of 21.32 kg, this would exclude many individuals from having the first two loads tested and result in too few data points to accurately calculate a line of best fit using a downward exponential association. We therefore adapted the protocol to utilising 30, 40, 50, 60 and 70% of maximum rebound bench press. However, this protocol was limited to individuals with a 1RM



rebound bench press of 70 kg and above. Individuals with a lesser 1RM would have to be excluded due to having too few data points to accurately measure their line of best fit.

In previous studies using this technique, the bar was lowered to  $\pm 3$  cm above the chest and held for  $\pm 0.6$  seconds before the perturbation was performed (155). We found that this specific angle was unstable and that the subjects we initially tested could not maintain most loads sufficiently during the perturbation. We therefore changed the position of holding the bar to a standardised  $90^\circ$  arm angle with the upper arms  $\pm 90^\circ$  abducted. We found this position to be more stable with more repeatable results (Figure 40).



**Figure 40:** *The positioning for the oscillation test of the upper extremities. Note also the Velcro-straps for stabilisation and position of perturbation.*

Before every load was measured, we again controlled the angle at which the bar was held to confirm the accuracy of measure. We found through trial and error that there were other factors that needed to be controlled. For example:

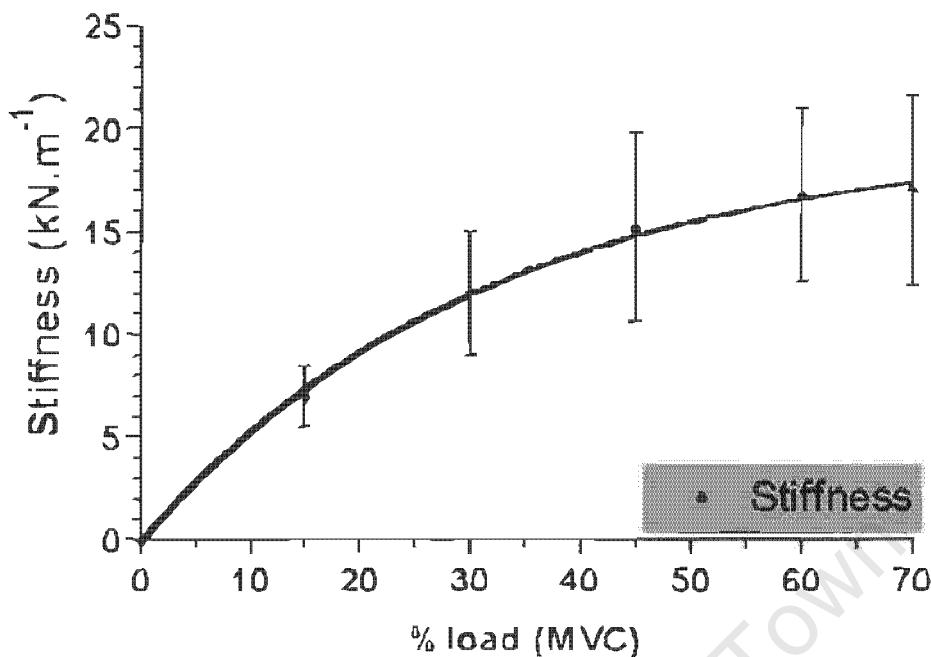
- The subject must lower the bar slowly to the position and hold still for  $\pm 0.6$  seconds before the perturbation – this removes any natural oscillation from occurring before the perturbation.
- The subject must maintain a deep breath and hold it to remain as still as possible during the holding and perturbation phases.

- The subject must close his/her eyes so that pre-activation before perturbation does not occur.
- The subject must maintain a force against the load, to prevent the muscles from giving way when the perturbation is applied.
- Verbal cues must be given to ensure that the subject follows the testing procedure carefully, as the end-result is dependent on the subject complying with the protocol.

Another prominent finding, was discovered during EMG testing. Irrespective of how soft the perturbation to the load was, there was always a small spike in the EMG after perturbation suggesting a small amount of additional muscle recruitment. Although, instructing the subject not to force any oscillations, but to let them occur naturally reduced this spike, the spike was always present. If the EMG significantly increased before (in anticipation) or increased drastically after (in response) the perturbation, the trial was not used in the calculation of stiffness. Furthermore, if the subjects voluntarily induced oscillations, as confirmed by coinciding rhythmic spikes in EMG, the trial was voided and repeated.

After we were satisfied with the set-up positioning and control measures, three subjects underwent preliminary testing. After determining their 1RM rebound bench press, the tests were performed. Each subject had six trials at each load with adequate rest periods between loads. After calculating the stiffness values for the six trials at every load, we adopted a standard protocol, which was used for all the studies. In accordance with this protocol the highest and the lowest stiffness values per load were excluded from the subsequent computation. If the value was more than double the estimated average at the specific load or less than half the estimated average, it was also excluded from further analysis. The remaining values, which in most cases was four, were averaged and the standard deviation calculated. The averaged value and standard deviation at each load were entered into Graphpad Prism 3.0 (Graphpad software Inc., San Diego, CA, U.S.A.) software and fitted to a downward exponential association curve fit and the line of best fit was determined (Figure 41). After the equation of best fit was determined, the stiffness at each load, and more importantly the maximal stiffness at 100% of 1RM, was calculated accordingly.

## Stiffness-load curve



**Figure 41:** The downward exponential association best line of fit curve for maximal stiffness as determined in Graphpad Prism 3.0.

To ascertain the test re-test repeatability of this system, we repeated the tests, under controlled conditions, as previously explained. We plotted the predicted stiffness at all the loads of the first test against the predicted stiffness at all loads of the second test. We found satisfactory repeatability on all three subjects and therefore decided we were ready to start a more formal repeatability study.

### **THE OSCILLATION TEST OF THE LOWER BODY EXTREMITIES**

A similar procedure to the oscillation test of the upper body extremities was performed to standardise the procedure and set-up positioning for the lower body. Our system is unique and different to the system used by Walshe et al. (151), therefore we had to test the system in various positional arrangements. We tested various subjects at different angle settings and finally decided upon a knee angle of  $\pm 95^\circ$  during contraction as the set-up position for the oscillation test of the lower body extremities. The seat angle was standardised at  $155^\circ$ . The set-up and positioning were identical to the MVC leg press test discussed in the following section. These respective angles and the set angles during a test were controlled using the goniometer.

Once the required load was set, the tester removed the pull-pin and the subject was instructed to push, take the load and straighten the legs until the knees were locked. The subject was then instructed to load the muscles of the lower extremities by bending the knees until they were  $\pm 120^\circ$ . We found that this angle was the most stable angle to measure the oscillations. If the subjects bent their legs further, the majority could not maintain most loads sufficiently and therefore oscillations could not be measured accurately. If the angle was increased any further, the muscles were not loaded sufficiently.

Various control mechanisms were utilised so that the subject performed the test properly. We strapped the subject down at the hips using Velcro-straps (Figure 32) and positioned him/her so that the shoulders were tight against the shoulder pads of the leg press seat-section (Figure 31). The subject was initially adjusted to  $\pm 120^\circ$  using the goniometer as a reference, and then instructed to orientate him-/herself with respect to this angle. They were then asked to accommodate the load to the specified angle on their own after which this angle was confirmed. Once the subject could perform this angle estimation accurately on a regular basis, we started testing. At the beginning of each load, the angle was again confirmed. The subject was instructed to take the load to the specified angle, to hold and to maintain the load.

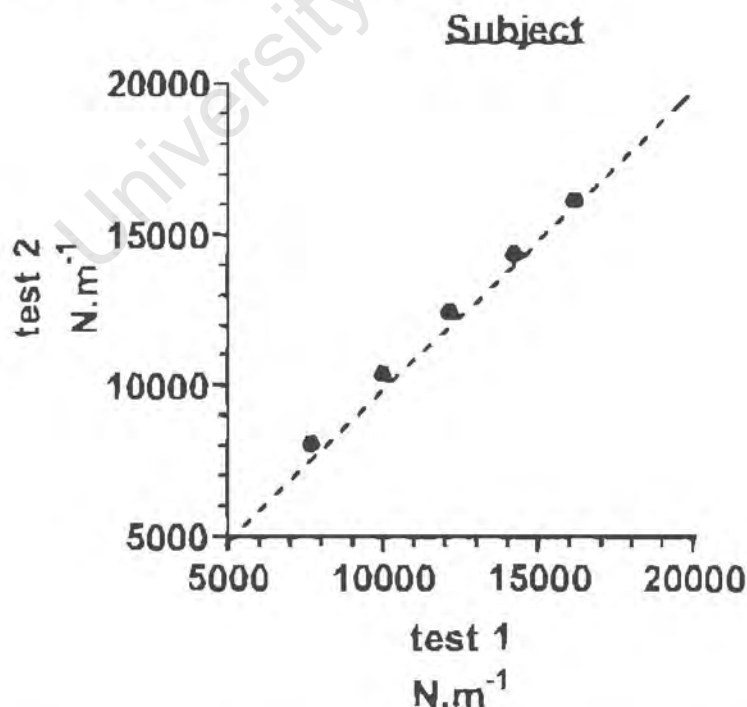
During the holding phase, the subject was asked to take a deep breath and close their eyes. This was then followed by a perturbation of between 50 and 200 Newton (Figure 42). The rationale behind the breath holding and closing of the eyes was to limit the amount of extraneous movement and to remove any pre-activation of the muscles before the perturbation of the load. The subject was further instructed not to respond to the perturbation and to only concentrate on maintaining a steady push against the load. The loads used were 30, 40, 50, 60 and 70% of maximum isometric leg press, which will be discussed in the following section. We tested the equipment until we were certain of the procedure running smoothly and then recruited a few subjects for preliminary tests.





**Figure 42:** This figure shows how the perturbation was performed on the loading arms of the leg press unit.

The same procedure for analysis and exclusion criteria were used for the lower body test, as previously described for the upper body test. After we performed repeated tests on three subjects, we plotted their initial test results versus their second test results to ascertain the linearity of the relationship. An example of one of the subjects' data collected during pilot testing is shown in Figure 43.



**Figure 43:** Graphical plot showing the repeatability of stiffness measurements between two trials for one of the subjects used in pilot testing.



We were satisfied with the results of our three subjects and decided that it was ready for a more formal analysis of repeatability.

### ***MVC LEG PRESS TEST***

The knee angle was the major determining factor in the test. We tested at different angles ranging from 90 - 135° for the MVC leg press test. Based on practicality of the measure and the load the system and subject could handle we decided on a similar knee angle ( $\pm 95^\circ$ ) to that used by Walshe and Wilson (149), as a reference for the set-up position for the maximum isometric leg press test. This was the same as the subject set-up for the oscillation test of the lower body extremities. The seat angle was standardised at 155°. These angles were also controlled using the goniometer.

To standardise the positioning for this test, the subject was strapped down and set-up using the goniometer in a similar way to the method described for the oscillation test. The subject, on command, was instructed to push as fast, and as hard as possible for three to five seconds. The best of three trials was used as the reference for determining the loads used for the oscillation test of the lower body extremities.

### ***MVC BENCH PRESS TEST***

For this test, the bench press bench was bolted down and the same set-up, control measures and precautions as described for the 1RM rebound bench press test were used. The upper stoppers of the modified Smith machine (Figure 7) were set so that when pressed upwards, the bar was pressed against them and fixed  $\pm 6$  cm above the chest height. The subject was instructed to press the bar lightly against the stoppers to remove impact artefact and then on command, to push as fast, and as hard as possible, upwards against the upper stoppers for three to five seconds. The best of three trials was used for further analysis.

### ***THE ISOLATED CONCENTRIC SQUAT JUMP TEST***

For this test there were various control measures that we had to standardise to increase the accuracy, reliability and repeatability of the measurement. The way the subject stands,

the foot positioning, the way that he/she holds the bar on the shoulders, and the angle of maximum depth were a few of the considerations that could influence the measurement.

After numerous trials, we standardised the testing procedure with the following guidelines:

- The bar must be centred on the base of the neck, i.e. supported on the trapezius and deltoid muscles
- The subject must grip the bar as close as possible to the shoulders, thereby naturally pulling the bar tight against the body.
- When standing the subject must stand with the legs hip width apart, because at landing the legs will be hip width apart and extended.
- The feet must be positioned directly underneath the body.
- When standing, the body must be aligned in the anatomical position of the frontal plane with the housing of the linear bearing bar-tracking unit. In other words, looking at the body from the side view, the line of the housing must run through the shoulder joint centre, through the hip joint centre, slightly anterior to the knee joint centre and anterior to the lateral malleolus of the ankle i.e. through the bridge of foot.
- The subject was instructed to bend the knees and squat down to a depth coinciding with a knee angle of  $\pm 95^\circ$ . This angle was controlled using a goniometer. The lower sound trigger was then set at this angle.
- The subject was then asked to bend further until  $\pm 90^\circ$ . Another marker was set on the housing of the linear bearing unit.

Once all these controls had been performed, the subject was reminded to keep the bar tight against the shoulders and to aim for maximum height when jumping (Figure 44).

For the actual test, the subject was instructed to squat until the specified  $\pm 90^\circ$  was achieved. He/she had to hold this angle stationary for  $\pm 0.5$  second, and then on command jumped as high as possible without releasing the bar. It was important that there was no countermovement. If any countermovement occurred, that trial had to be repeated. The best of three trials was used for further analysis.

## **THE COUNTERMOVEMENT JUMP TEST**

The same control measures were used as described in the methods for the squat jump test. However, during the squat jump test the sound trigger was not switched on. For this test, the sound trigger is switched on and the subject starts from an upright standing position. The subjects had to drop down as fast as possible and when they heard the sound trigger reverse their direction of movement and jump as high as possible. The trigger was set at  $\pm 95^\circ$ , so by the time they reversed direction they should have reached  $\pm 90^\circ$ . This method, controls the depth of the starting position for both the squat and countermovement jumps and ensures that their range of motion are the same. The logic behind this control of the squat and countermovement jumps was to make the comparison between the two conditions more interpretable.

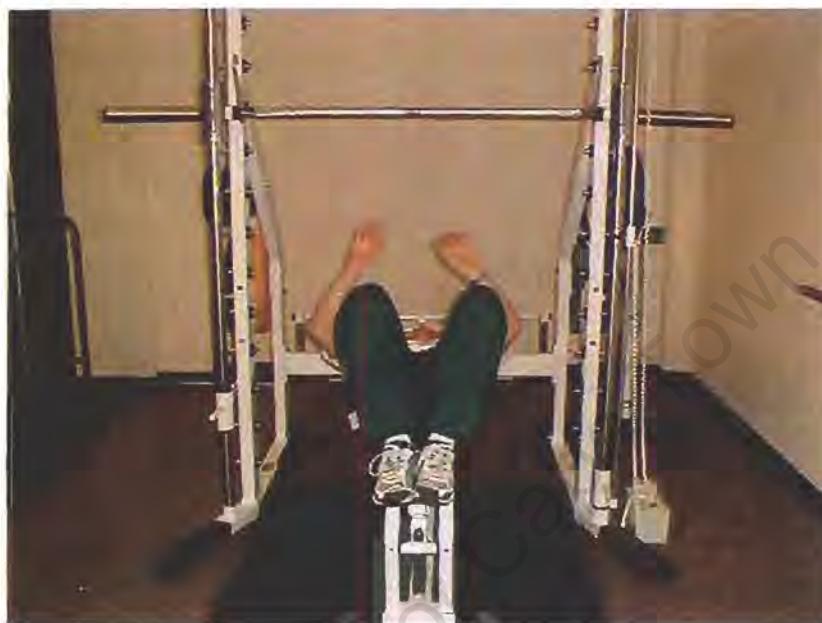


**Figure 44:** A front view of a subject during the jumping phase of a squat jump

## **THE PURE CONCENTRIC BENCH THROW TEST**

The same set-up and positioning as described for both the 1RM rebound bench press and the oscillation test for the upper body extremities was performed during the pure concentric bench throws (Figure 45). The sound trigger was also set up as for the 1RM rebound bench press test. However, another marker was set on the housing of the linear

bearing tracking-unit, which indicates the lower position of  $\pm 3$  cm above the chest. For this test, the subject was instructed to lower the bar to the set position, to hold for  $\pm 1$  second and then on command to push and throw the bar as high as possible. The subject was also instructed to catch the bar with straight arms, before absorbing the load. The normal safety precautions i.e. the lower stoppers to prevent injury in the event of missing the bar or not being able to control it down were included. Once again, if there was evidence of any countermovement, the trial was not used and had to be repeated.



**Figure 45:** A subject after just having released the bar during a pure concentric bench throw.

### **THE REBOUND BENCH THROW TEST**

The lower sound trigger was set up as previously described, to standardise the lower position of the 1RM rebound bench throw test. The sound trigger was set to signal reversal of direction when the bar was  $\pm 6$  cm from the chest. This gave the subject enough time to react and reverse the direction of push within  $\pm 3$  cm. The subject had to quickly lower the bar until the sound trigger was heard. He/she then had to reverse the direction of movement, push and throw the bar as high as possible before catching it again as previously described.

## **SUMMARY**

This chapter describes in a step-by-step way, how the equipment was designed, built and then also how the protocols were developed and subjected to preliminary testing. After all these procedures and techniques were standardised, practiced and finalised, the next phase of research began in which the repeatability of these measurements were evaluated in a more systematic way. The following three chapters are focused on the methodology, repeatability and refinement of the final testing procedures defined in this chapter.

University of Cape Town



## **CHAPTER 3:**

# **The repeatability of the oscillation test for determination of musculotendinous stiffness in both upper and lower body using the NAMS unit**

## **INTRODUCTION**

Based on the literature reviewed in Chapter 1, it is clear that muscle and tendon elasticity has an important role to play in regulating a muscle's contractile properties (48;97;132;136;149;155;156;159). In particular, stretch shortening cycle performance is possibly influenced by the stiffness of the muscle-tendon complex (MTC) (33;132;136;149;153;155), and by altering the tension of the muscle before contraction. There are divergent views regarding the relationship between stiffness of the muscle-tendon complex and stretch shortening cycle muscle function. For example, some researchers suggest that a more compliant muscle-tendon complex is more advantageous (7;33;132;156), whereas others suggest that a stiffer muscle-tendon complex (9;115;136) is more advantageous for stretch shortening cycle performance. Therefore, the relationship between these factors during stretch shortening cycle performance is not completely understood.

However, much of the variance in the literature can be attributed to the difficulty of measuring the *in vivo* stiffness of the muscle-tendon complex. It is logical that before musculotendinous stiffness can be studied and related to other variables it needs to be measured accurately. Therefore, the purpose of this study was to test the repeatability of the testing procedures for measuring stiffness of the muscle-tendon complex using the newly designed NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town, Cape Town, South Africa) discussed in Chapter 2.

# THE LOWER BODY

## METHODS

### SUBJECTS

10 Male subjects, between the ages of 20 and 31 years and with various training backgrounds were recruited for this study. Subjects were required to have resistance training experience of at least one year. Before subjects were tested for the main study, a screening test was performed to ascertain whether each subject could perform the test properly. A decision was made to include the screening test, because during preliminary testing on the apparatus, it was found that a few subjects could not master the technique of supporting the load with the constant force output necessary to maintain static equilibrium. The subjects who passed the screening test ( $n = 10$ ) were further tested on four different occasions. Their general characteristics are summarized in Table 1.

Body fat was measured and expressed as a % (43) and as a sum of 7 skinfolds (129).

**Table 1:** *Personal characteristics of subject group ( $n = 10$ ).*

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)	MVC (kg)(*)
25.8 $\pm 3.6$	175.8 $\pm 6.2$	78.8 $\pm 13.7$	75.1 $\pm 18.0$	15.6 $\pm 2.7$	66.2 $\pm 10.3$	290 $\pm 77$

(\*) Maximum Voluntary Contraction

All subjects were informed about the study and the potential risks involved. The Ethics Committee of the University of Cape Town approved the study. Each subject completed a personal questionnaire regarding medical, training and general information and further signed a written informed consent form complying with the guidelines set by the American College of Sports Medicine, before participating in the study (3).

### EXPERIMENTAL DESIGN

The first day was a familiarization day, in which the complete testing procedure was performed as a trial run. This was performed to ensure that the subject knew what was



expected of him, and to enhance the reliability of the testing procedure. After the familiarization test, the subjects had to visit the laboratory on a further three occasions where the tests were repeated. The subjects were instructed to wear the same shoes on all the testing days.

To ensure that the subjects had fully recovered from the previous day's testing, a 48-hour break was given between testing days. Subjects were asked to refrain from any lower body training at least 24 hours before testing and to refrain from any heavy or intense training throughout the duration of the study.

On the first day body composition and maximal voluntary contraction (MVC) or isometric leg press were measured. After these tests, the subjects were given the familiarization session, where the various oscillation technique procedures performed during the screening test were again demonstrated. The subjects were given between 3 and 6 trials per load to perfect the technique and become familiar with the procedure and what was expected from them.

The second testing day started with testing for maximal voluntary contraction (MVC) using the leg press unit. This was followed by the maximal muscle-tendon complex stiffness test for the lower body. On the third and fourth testing days, the maximal muscle-tendon complex stiffness test was repeated. Each day's testing was preceded by a standardized warm-up of 10 minutes of continuous, self-paced, submaximal shuttle-runs, specific stretches for the lower body and 2 sets of 10 repetition full squats with 20 kg (zero load on the bar).

#### **TESTING FOR THE MVC ISOMETRIC LEG PRESS**

For the lower body, the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town, Cape Town, South Africa) was set up and the subject positioned as previously described in Chapter 2 (p 124). This position was used to simulate a functional measure of MVC using muscle groups in a similar position to that used during jumping. Each subject was given 3 submaximal warm-up trials, where they were instructed to push at an estimated 50, 70 and 90% of maximum respectively against the immovable resistance of the stiffness system. Three maximal trials per subject followed the warm-up, with 2 - 3 minutes rest between trials. The maximum isometric force obtained from the three trials was recorded and used to determine the loads for the lower body stiffness test. The

subjects were instructed to push as fast and as hard as possible, maintaining the contraction for ~5 seconds against the isometric resistance. To ensure that the subjects exerted maximal effort in the MVC, the researchers gave vocal encouragement.

#### THE OSCILLATION TECHNIQUE

The oscillation technique was used for the measurement of MTC stiffness (32;132;149;156). This technique measures the fluctuation or oscillation in force with a loaded muscle group in a semi-static or stationary position when an additional external force is applied to the system (149;156). The fluctuation or oscillation in force registration via the load cell was used to calculate the stiffness of the muscle-tendon complex. Surface EMG was measured to eliminate those tests that had noticeable pre-activation as this neural intervention could influence the stiffness measures (149;156).

For the lower body oscillation test the procedure described in Chapter 2 (p 121 - 123) was used and subjects were instructed to maintain a constant muscular activity and force and not to respond to the perturbation in any way (106;156).

The force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

The resultant damped oscillations were modelled on a damped mass-spring model to determine the elastic stiffness at each load (132). The resulting damped oscillation was modelled by the following equation (132;149;151-153;155;156):

$$m(d^2x/dt^2)+c^{dx}/dt+kx = mg \quad (a)$$

Where:

- **x** = position of the involved structure (e.g. muscle-tendon complex),
- **g** = acceleration of gravity,
- **m** = mass of the system (e.g. leg press unit and the added weights) and
- **k** = elastic stiffness (e.g. of the muscle-tendon complex).

Using this formula, the force generated is therefore a function of the displacement, the damping and elastic stiffness of the involved structure (e.g. muscle-tendon complex).

Stiffness ( $k$ ) was determined by the equation (32;80;112;149;151;153;155;156):

$$k = 4(mf^2\pi^2) + (c^2/4m) \quad (b)$$

Where:

- $m$  = mass of the system,
- $f$  = damped natural frequency, and
- $c$  = the damping coefficient.

The damped natural frequency ( $f$ ) is calculated as the inverse of the time-period between two force-peaks ( $1/T$ ) or the natural frequency of oscillation of the damped mass spring unit, i.e. muscle-tendon complex (80).

After calculating the natural log of peak forces, the damping ratio  $s$  was calculated (39;80):

$$s = [\ln(F_1) - \ln(F_2)] / \{[\ln(F_1) - \ln(F_2)]^2 + 4\pi^2\}^{1/2} \quad (c)$$

Where:

- $\ln$  = natural log
- $F_1$  and  $F_2$  = force peaks

The following equation was used to determine the damping coefficient  $c$  (149;151;153;155;156):

$$c = 4\pi mf_n s \quad (d)$$

Where natural frequency ( $f_n$ ) was calculated using the equation (149;151;153;155;156):

$$f_n = \{f^2 / (1 - s^2)\}^{1/2} \quad (e)$$

Each subject was tested at various percentages of maximum isometric leg press, i.e. 30, 40, 50, 60, and 70 %. Six measurements were recorded at each load using the procedure outlined in Chapter 2 (p 121 - 123). Four of these measurements were used and averaged to determine the respective stiffness at each load. A rest of 2 - 3 minutes was allowed between each load and  $\pm 1$  minute between trials. Additional rest was given if required. The stiffness values per load were then plotted on a graph using the Graphpad Prism Version 3.0 software (Graphpad Software Inc., San Diego, California, USA) and the line of best fit determined using a downward exponential association and least sum of squares criterion.

Maximal stiffness was determined using the specific curve fit of the stiffness-load curve, using the equation:

$$\text{Stiffness} = Y\text{-max} \times (1 - \text{Exp}^{(-K \times X)}) \quad (f)$$

Where:

- **Y-max** = maximum plateau,
- **k** = a constant and
- **X** = percentage load.

Maximal stiffness was calculated from an extrapolated maximum load using the equation of the specific curve fit of the stiffness-load curve. This procedure has been recommended because most subjects cannot maintain stability at higher loads (149;151;153;155;156). Maximal stiffness calculated this way was assumed to be representative of the averaged stiffness of the tendons of the muscles of the lower body (149;151).

#### **ELECTROMYOGRAPHY (EMG)**

A potential problem measuring stiffness with the oscillation technique is that of pre-activation and reflex-facilitation. Surface EMG was therefore measured before and after the perturbation using LabVIEW 6.0.2 (National Instruments, Austin, Texas, USA) software and a telemetric EMG system (Noraxon, USA, Inc) to identify trials, which were confounded by neurological intervention (149;151;155;156).

The vastus medialis oblique muscle was used as a control measure for neural intervention. Two surface EMG electrodes (blue sensor SP-00-S, Medicotest A/S, Rugmarken, Denmark) were attached as a pair over the muscle belly with an inter-electrode distance of ~2 cm (135). The site for electrode placement was determined as the center of the muscle belly when contracted isometrically, and the electrodes were attached running parallel to the muscle fibers. A third neutral reference electrode was placed on the anterior tibia (82). Electromechanical delay was treated as a systematic error, as it was assumed that this was either negligible or constant (60).

Before the electrodes were attached, the skin surface was prepared. Hair was shaved off using a razor and the skin was scraped with sandpaper to remove the outer layer of epidermal skin cells. The skin surface was then swabbed clean, removing any oils or dirt, using an alcohol swab. Once the alcohol had evaporated, the electrodes were placed on the skin.

The EMG was observed on-screen as an indicator of excessive pre-activation. If the EMG measure remained relatively constant before the perturbation it was assumed that the stiffness measure was the result of the elasticity of the series elastic components of the muscle-tendon complex (149;151). If the EMG increased before the perturbation, it was presumed that pre-activation in anticipation of the perturbation had occurred and the trial was excluded and redone. This oscillation test procedure was repeated as described above on all the testing days.

## **STATISTICAL ANALYSIS**

The data were analysed using Statistica '99 Edition statistical analysis software (Statsoft Inc., Tulsa, U.S.A.). The data were expressed as the mean  $\pm$  standard deviation (SD). The intra-class correlation coefficient was calculated for the stiffness measures obtained on each day's testing. The intra-class correlation coefficient was used to define the repeated measures data on the same variable (8). According to Vincent (146) an intra-class correlation coefficient of  $R = 0.7 - 0.8$  is considered to be of questionable reliability,  $R = 0.8 - 0.9$  of moderate reliability and  $R = 0.9$  or greater is considered as being very high and shows a consistency of measurement across trials. These values should, however, be considered as guidelines as they were not derived from experimentation (8).

It seems as if the intra-class correlation coefficient is prone to individual variance and is affected by sample heterogeneity to such a degree that a high correlation may still mean an unacceptable error and therefore should not be the sole statistic used for reliability studies (8). The coefficient of variation ( $(SD/X) \times 100$ ) was therefore also calculated for each individual's stiffness variables and averaged (8). On the basis of the findings of Atkinson and Nevill (8) and the approach adopted by Morrow and Jackson (117), the 95% confidence intervals were then calculated for the intra-class correlation coefficient and coefficient of variation using online available software (76). This approach provides a range of values that express the "true" reliability range (117).

A one-way analysis of variance (ANOVA) for repeated measures was then performed on the data to further assess the reliability of these measures. As the one-way ANOVA was performed on the same group of individuals throughout each test, inter-individual variability was eliminated. An important assumption specific to repeated measure designs of this nature is that of compound symmetry or sphericity (157). This assumption requires that the variances (within group) and covariances (across subjects) of the repeated measures are homogenous or identical. If this assumption is violated, i.e. there were differences between the three trials, the F-ratios and F-values will be inflated and the likelihood of making a type 1 error is enhanced (157).

Mauchley's test for sphericity was used to ascertain if there was a significant difference between trials (157). If this test result was not significant, it was assumed that there was sphericity among trials and no adjustments needed to be made. If the result was significant, it was assumed that sphericity was not met. In this case Greenhouse/Geiser and Huynh-Feldt adjustments (157) to the degrees of freedom were made by the software to counter the increased risk of making a type 1 error (i.e. detecting differences, which were not real). If the F-value was still significant after this adjustment the null hypothesis was rejected (157) and a Scheffe's post-hoc analysis was performed to identify the differences between trials. The Scheffe's post-hoc test was chosen as it is a relatively conservative test and guards against making a type 1 error i.e. showing differences where there are not. The T-test for independent variables was used to measure differences between data sets.

The data from the second day were additionally compared to the data of the third testing day using the method of Bland-Altman (16) to calculate the precision of measurement between these two tests. This method measures the limits of agreement between the two data sets by plotting the standard deviation of their difference against the average measure of the two tests. The precision is then determined by calculating two standard deviations above and below the average difference (16). The closer the average difference is to zero, and the smaller the range above and below zero, the better the precision. It is also assumed that the majority of the data should fall within two standard deviations above and below the average difference (16).

## RESULTS

Table 2 shows the average stiffness measure for all three trials at all the relative loads for each individual.

**Table 2:** Average stiffness data ( $\text{kN.m}^{-1}$ ) at each relative load for each individual between three trials ( $\bar{X} \pm \text{SD}$ ) ( $n = 10$ ). CV100% represents the coefficient of variation of maximal stiffness.

SUBJECT	30%	40%	50%	60%	70%	100%	CV100%
Lower 1	13.5 $\pm$ 0.6	16.8 $\pm$ 0.7	19.6 $\pm$ 0.6	22.1 $\pm$ 0.6	24.2 $\pm$ 0.5	29.0 $\pm$ 0.3	1.2
Lower 2	9.1 $\pm$ 1.5	11.9 $\pm$ 1.7	14.6 $\pm$ 1.6	17.2 $\pm$ 1.4	19.8 $\pm$ 1.1	27.0 $\pm$ 0.9	3.4
Lower 3	12.9 $\pm$ 1.5	16.0 $\pm$ 1.6	18.8 $\pm$ 1.7	21.1 $\pm$ 1.6	23.1 $\pm$ 1.5	27.7 $\pm$ 1.1	4.0
Lower 4	17.0 $\pm$ 0.7	21.0 $\pm$ 0.8	24.3 $\pm$ 0.5	27.2 $\pm$ 0.4	29.6 $\pm$ 0.2	34.8 $\pm$ 0.5	1.4
Lower 5	18.0 $\pm$ 0.5	21.9 $\pm$ 0.6	25.0 $\pm$ 0.7	27.6 $\pm$ 0.9	29.7 $\pm$ 1.0	33.8 $\pm$ 1.4	4.0
Lower 6	17.1 $\pm$ 0.9	21.4 $\pm$ 0.7	25.1 $\pm$ 0.5	28.4 $\pm$ 0.5	31.2 $\pm$ 0.7	37.7 $\pm$ 2.2	5.8
Lower 7	17.3 $\pm$ 2.0	20.7 $\pm$ 1.6	23.5 $\pm$ 0.9	25.7 $\pm$ 0.1	27.4 $\pm$ 0.9	31.1 $\pm$ 3.6	11.6
Lower 8	6.0 $\pm$ 0.4	7.9 $\pm$ 0.5	9.6 $\pm$ 0.6	11.3 $\pm$ 0.8	13.0 $\pm$ 1.0	17.5 $\pm$ 1.9	11.3
Lower 9	22.0 $\pm$ 1.3	26.0 $\pm$ 1.0	29.2 $\pm$ 0.6	31.5 $\pm$ 0.3	33.3 $\pm$ 0.3	36.5 $\pm$ 1.2	3.3
Lower 10	19.0 $\pm$ 2.1	22.6 $\pm$ 1.8	25.3 $\pm$ 1.4	27.4 $\pm$ 0.9	29.0 $\pm$ 0.6	31.9 $\pm$ 1.1	3.5
Mean	15.2 $\pm$ 4.8	18.6 $\pm$ 4.8	21.5 $\pm$ 5.9	23.9 $\pm$ 6.1	26.0 $\pm$ 6.1	30.7 $\pm$ 5.9	4.9 $\pm$ 3.7

Table 3 shows the intra-class correlation coefficient and coefficient of variation together with their respective 95% confidence intervals. The intra-class correlation coefficient and the average coefficient of variation for maximal stiffness was  $R = 0.97$  (95% CI: 0.92-0.99) and  $CV = 4.9$  (95% CI: 3.7-7.3)% respectively.



**Table 3:** Repeatability measures of musculotendinous stiffness of the lower extremities (n = 10).

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
MTC stiffness@30%MVC	0.98 (0.94-0.99)	8.0 (6.0-11.8)
MTC stiffness@40%MVC	0.98 (0.94-0.99)	6.4 (4.8-9.5)
MTC stiffness@50%MVC	0.99 (0.97-1.00)	4.8 (3.6-7.1)
MTC stiffness@60%MVC	0.99 (0.97-1.00)	3.6 (2.7-5.3)
MTC stiffness@70%MVC	0.99 (0.97-1.00)	3.4 (2.6-5.0)
MTC stiffness@100%MVC	0.97 (0.92-0.99)	4.9 (3.7-7.3)

**Index:** MTC = Muscle-tendon complex, MVC = maximal voluntary contraction

Table 4 shows that the 70% stiffness values were significantly different over the three trials. The Scheffe's post-hoc test was then performed and the findings are represented in Table 5.

**Table 4:** One-way analysis of variance in the lower body musculotendinous stiffness test (n = 10).

Variable	df effect	MS effect	Df error	MS error	F	p-level (*significant)
MTC@30%MVC	2	1.07	18	1.73	0.62	0.55
MTC@40%MVC	2	1.12	18	1.46	0.77	0.48
MTC@50%MVC	2	1.25	18	1.0	1.26	0.31
MTC@60%MVC	2	1.70	18	0.66	2.59	0.10
MTC@70%MVC	2	2.27	18	0.59	3.84	0.04*
MTC@100%MVC	2	3.49	18	2.78	1.25	0.31

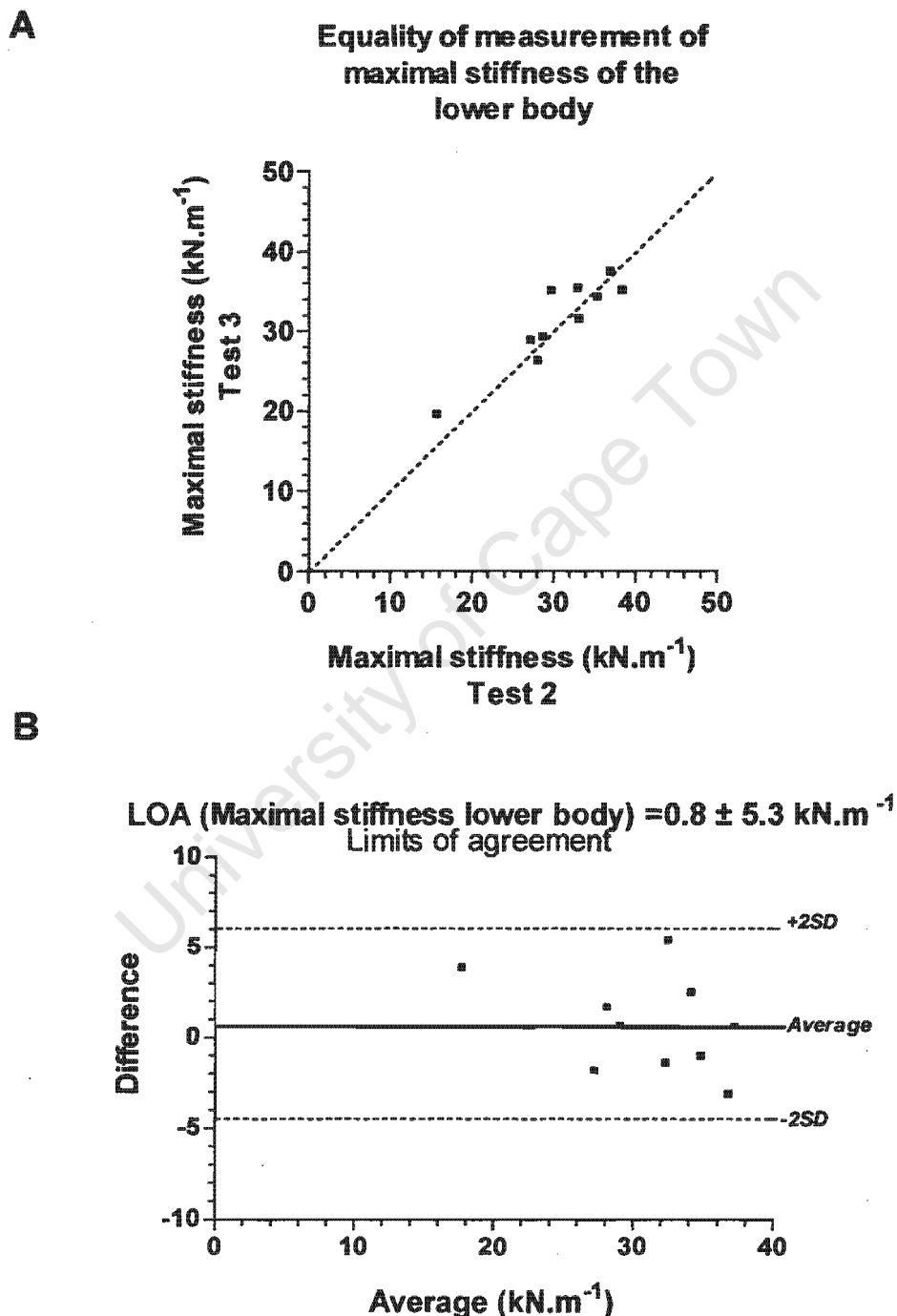
**Index:** MTC = Muscle-tendon complex, MVC = maximal voluntary contraction

**Table 5:** Scheffe's Post-hoc test for differences between trials in the lower body musculotendinous stiffness test (n = 10).

Variable: MTC stiffness @ 70% MVC (*significant)			
Testing day	Average day 1: 25.72	Average day 2: 25.78	Average day 3: 26.57
1		0.98	0.07
2	0.98		0.96
3	0.07	0.96	

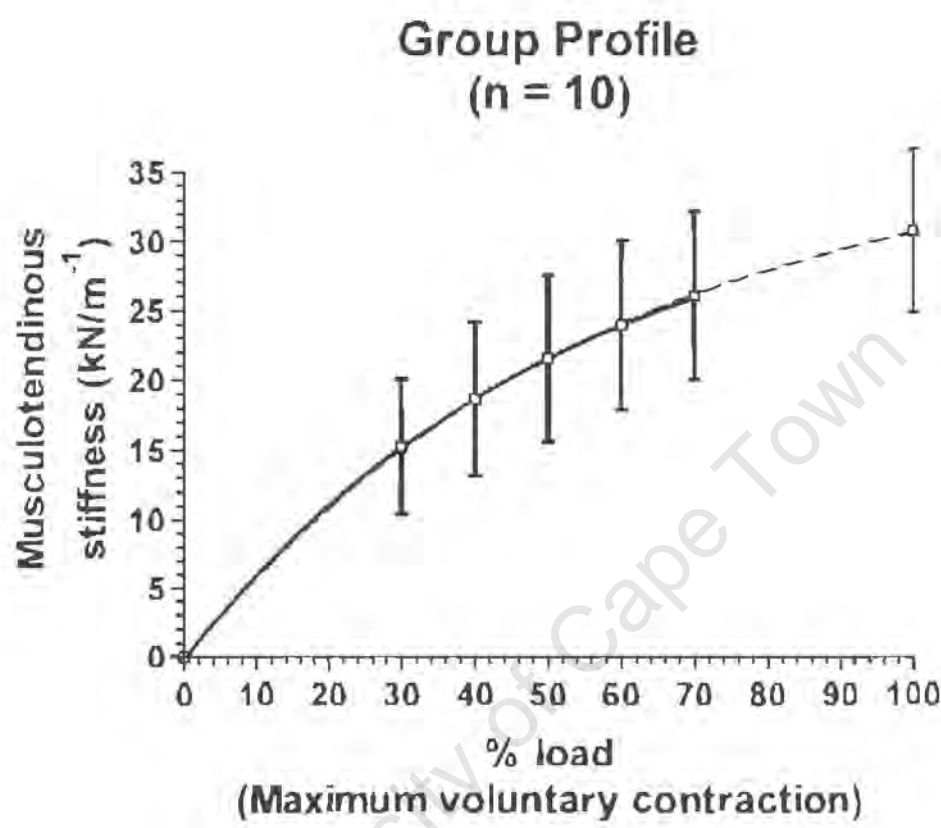
**Index:** MTC = Muscle-tendon complex, MVC = maximal voluntary contraction

Figure 1 A is a graph of the maximal stiffness (test 2 vs. test 3). Figure 1 B shows that the precision (two standard deviations above and below the average difference between trials) by limits of agreement, according to Bland-Altman (16) for measurement of maximal musculotendinous stiffness of the lower body falls within a range of  $10.6 \text{ kNm}^{-1}$  ( $0.8 \pm 5.3 \text{ kN.m}^{-1}$ ).



**Figure 1:** A - The equality of measurement for maximal stiffness between the second and third testing day using the lower body oscillation test. B – The limits of agreement for measurement of maximal stiffness between the second and third testing day using the method of Bland-Altman (16).

The general tendency of the stiffness-load curve of the series elastic elements of muscle and tendon followed the classic curvilinear model, which was also found in earlier studies (151;153;156) (Figure 2). The average maximal stiffness of the total group was  $30.7 \pm 5.9 \text{ kN.m}^{-1}$ .



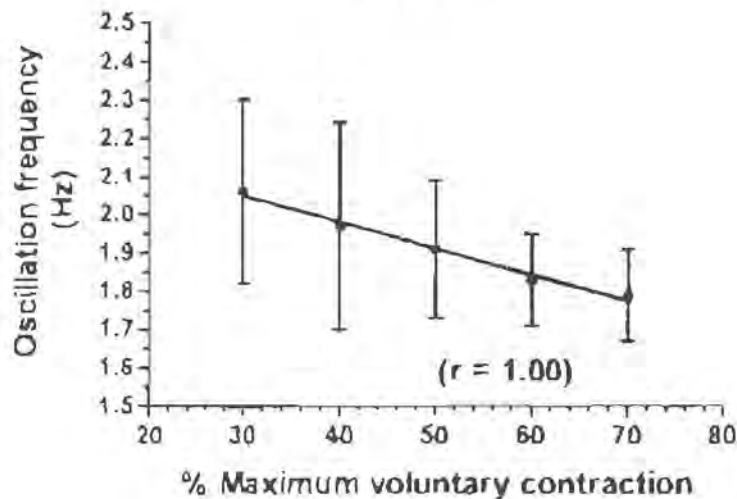
**Figure 2:** The curvilinear relationship between musculoskeletal stiffness and load (n = 10). Values are expressed as  $\bar{X} \pm \text{SD}$ .

The frequency of oscillation decreased as the load on the muscle-tendon complex increased (Table 6, Figure 3). The mean damped natural frequency of oscillation varied between 1.79 and 2.06 Hz over the various loading conditions (Table 6).

**Table 6:** The average damped natural frequency data at the relative loads using the NAMS unit (n = 10 x 6 trials per load).

% MVC	Damped Natural Frequency (Hz)
30%	2.06 ± 0.24
40%	1.97 ± 0.27
50%	1.91 ± 0.18
60%	1.83 ± 0.12
70%	1.79 ± 0.12

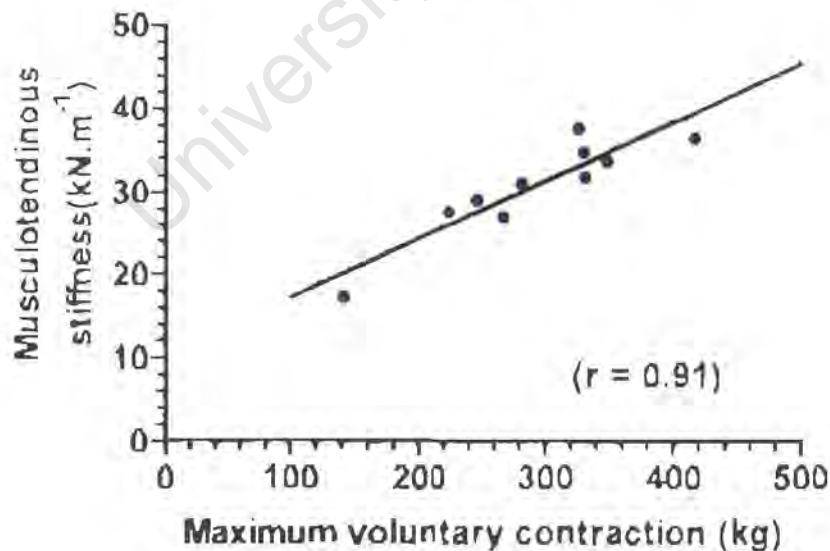
### Damped natural frequency of oscillation vs % MVC (n = 10)



**Figure 3:** Relationship between %MVC and oscillation frequency (Hz) ( $n = 10 \times 6$  trials per load).

There was a significant relationship ( $r = 0.91$ ,  $p < 0.0002$ ) between the MVC and maximal stiffness (Figure 4)

### Maximal stiffness vs MVC (n = 10)



**Figure 4:** Relationship between maximum voluntary contraction (MVC) and maximal musculotendinous stiffness ( $n = 10$ )

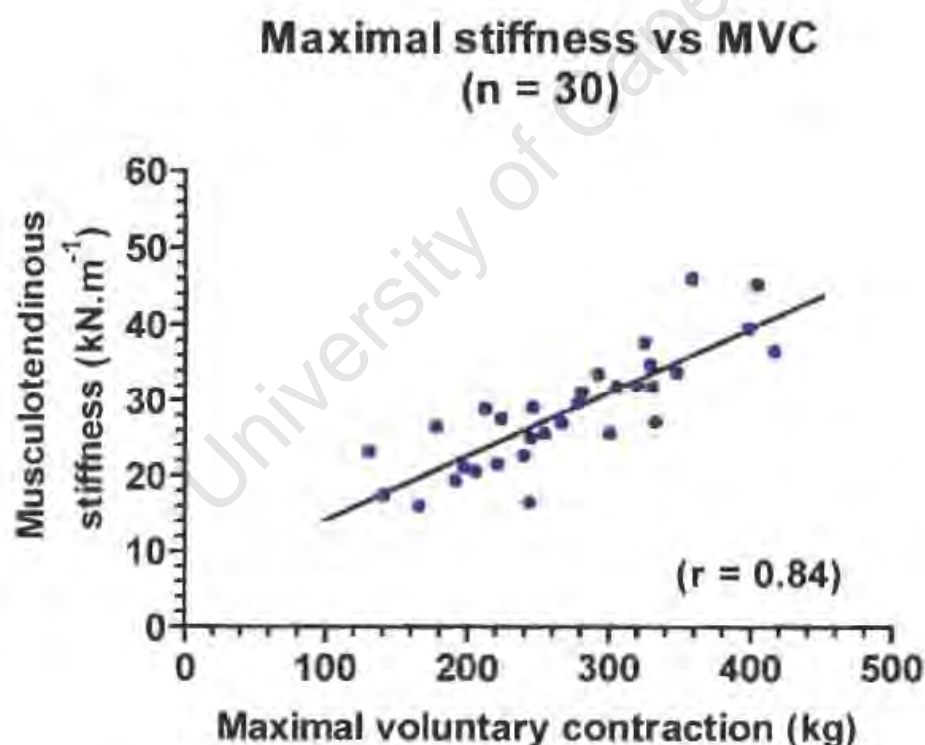
After demonstrating that the musculotendinous stiffness data were repeatable, and in an attempt to confirm, strengthen and solidify any relationships that were identified, we included data from an additional 20 subjects tested in earlier pilot studies (Tables 7 and 8) (71;143). Their combined personal characteristics are represented in the following table (Table 7).

**Table 7:** Personal characteristics of the enlarged subject group ( $n = 30$ ).

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)	MVC (kg)(*)
25.8 $\pm 3.4$	179.0 $\pm 7.6$	79.7 $\pm 11.7$	70.0 $\pm 15.9$	14.8 $\pm 2.6$	68.1 $\pm 9.3$	270 $\pm 9$

(\*) Maximum Voluntary Contraction

A similar relationship between maximal musculotendinous stiffness and MVC occurred for this enlarged group of subjects ( $r = 0.84$ ,  $p < 0.0001$ ) (Figure 5).



**Figure 5:** Relationship between maximum voluntary contraction (kg) and maximal musculotendinous stiffness ( $n = 30$ ).

Table 8 represents the combined larger group stiffness data ( $n = 30$ ) at all the relative loads for each individual.

**Table 8:** *Musculotendinous characteristics of the grouped subjects (n = 30).*

SUBJECT	30%	40%	50%	60%	70%	100%
JG	7.7	10.0	12.1	14.1	16.1	21.2
JJ	9.1	11.5	13.7	15.7	17.4	21.6
JGR	13.8	17.3	20.4	23.0	25.4	30.7
JR	10.3	12.7	14.8	16.5	17.9	21.1
JRO	7.5	9.3	11.0	12.4	13.7	16.5
JM	8.9	11.6	14.1	16.5	18.8	25.0
JGM	7.9	10.2	12.4	14.6	16.6	22.0
JA	10.5	12.7	14.5	16.0	17.1	19.4
JAN	12.0	15.5	19.0	22.2	25.3	33.5
JGB	7.4	9.7	11.8	13.8	15.8	21.1
JLE	16.9	21.8	26.4	30.7	34.7	45.2
JW	13.3	16.7	19.7	22.3	24.5	29.7
JB	18.2	22.7	26.6	30.0	32.9	39.5
JS	12.7	16.0	18.9	21.4	23.7	28.9
JP	18.5	23.7	28.3	32.5	36.4	46.0
CAT2	8.5	11.1	13.6	15.9	18.2	24.5
CAT3	9.1	11.8	14.4	16.9	19.2	25.6
CAT5	12.9	16.0	18.7	21.0	23.0	27.4
CAT7	15.6	20.0	24.0	27.7	31.1	39.7
CAT10	12.0	15.4	18.7	21.7	24.5	31.9
CAT11	11.2	14.0	16.5	18.6	20.5	24.9
CAT8	11.8	15.2	18.5	21.5	24.4	32.0
CAT12	5.8	7.5	9.1	10.7	12.1	16.0
CAT15	15.4	18.3	20.5	22.1	23.4	25.6
CAT6	9.1	11.4	13.4	15.3	16.8	20.6
CAT13	12.6	16.3	19.8	23.2	26.4	34.9
CAT16	8.1	10.5	12.9	15.1	17.3	23.2
CAT17	10.0	12.6	14.8	16.8	18.6	22.7
CAT18	15.6	18.7	21.1	23.0	24.5	27.2
CAT19	11.3	14.3	16.9	19.3	21.4	26.5
CAT20	14.7	18.2	21.1	23.5	25.6	30.0
Lower 1	13.5	16.8	19.6	22.1	24.2	29.0
Lower 2	9.1	11.9	14.6	17.2	19.8	27.0
Lower 3	12.9	16.1	18.8	21.1	23.1	27.6
Lower 4	17.0	21.0	24.3	27.2	29.6	34.8
Lower 5	18.0	21.9	25.1	27.6	29.7	33.8
Lower 6	17.1	21.4	25.1	28.4	31.2	37.7
Lower 7	17.3	20.7	23.5	25.6	27.4	31.1
Lower 8	6.0	7.9	9.6	11.3	13.0	17.5
Lower 9	22.0	26.1	29.2	31.5	33.3	36.5
Lower 10	19.1	22.6	25.3	27.4	29.0	31.9
Mean	12.4 ± 4.0	15.6 ± 4.8	18.4 ± 5.3	20.8 ± 5.8	23.0 ± 6.2	28.3 ± 7.3

The stiffness data were then arranged in ascending order and then two groups were formed after separating the 15 stiffest subjects from the remaining 15 subjects. The maximal stiffness values of the stiff group ( $34.8 \pm 5.4 \text{ kN.m}^{-1}$ ) were significantly ( $p < 0.0000001$ ) greater than the compliant group ( $22.4 \pm 3.8 \text{ kN.m}^{-1}$ ). A similar strategy of analysing data has been reported previously (149). The MVC for the stiffer group ( $322 \pm 57 \text{ kg}$ ) was significantly higher ( $p < 0.00004$ ) than the more compliant group ( $220 \pm 56 \text{ kg}$ ). This relationship between stiffness and %MVC was plotted for each group (Figure 6). It is clear from Figure 6 that the stiffer group had greater stiffness at each %MVC compared to the more compliant group.



### Stiff Vs Compliant subjects (n = 30)

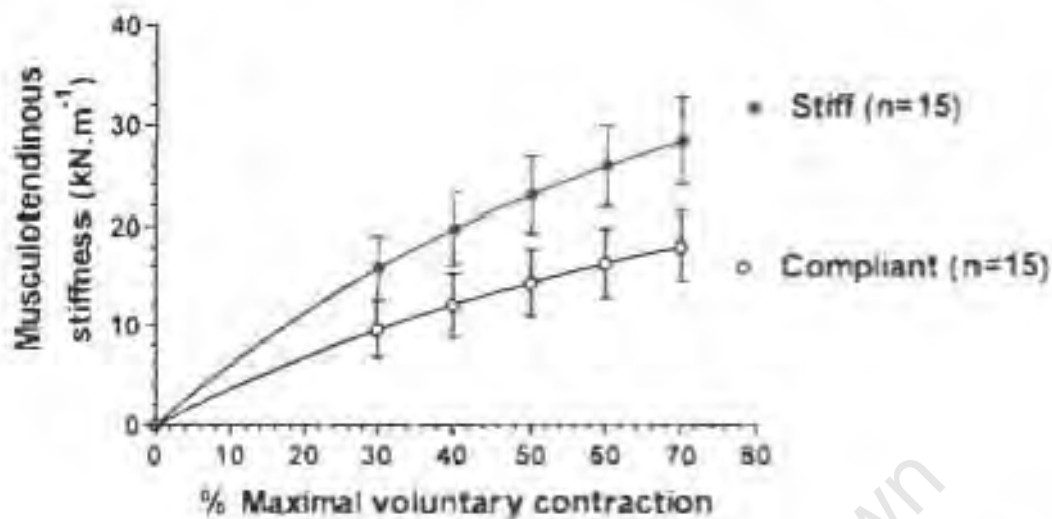


Figure 6: Relationship between stiffness (kN.m<sup>-1</sup>) and %MVC for stiff (n = 15) and compliant (n = 15) subjects.

The stiffer group had a significantly greater MVC than the compliant group.

After further analysis MVC was also significantly related to the body mass of the individual ( $r = 0.63$ ,  $p < 0.0002$ ) (Figure 7).

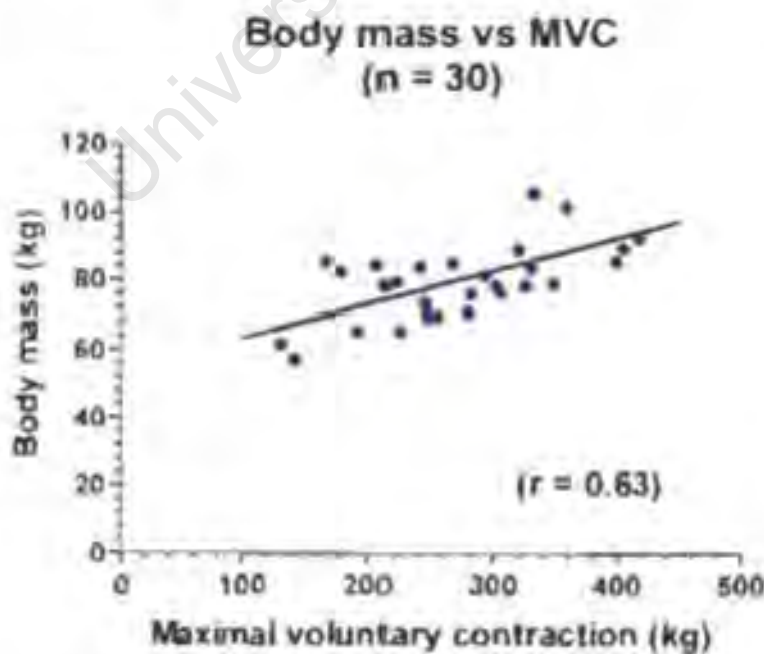
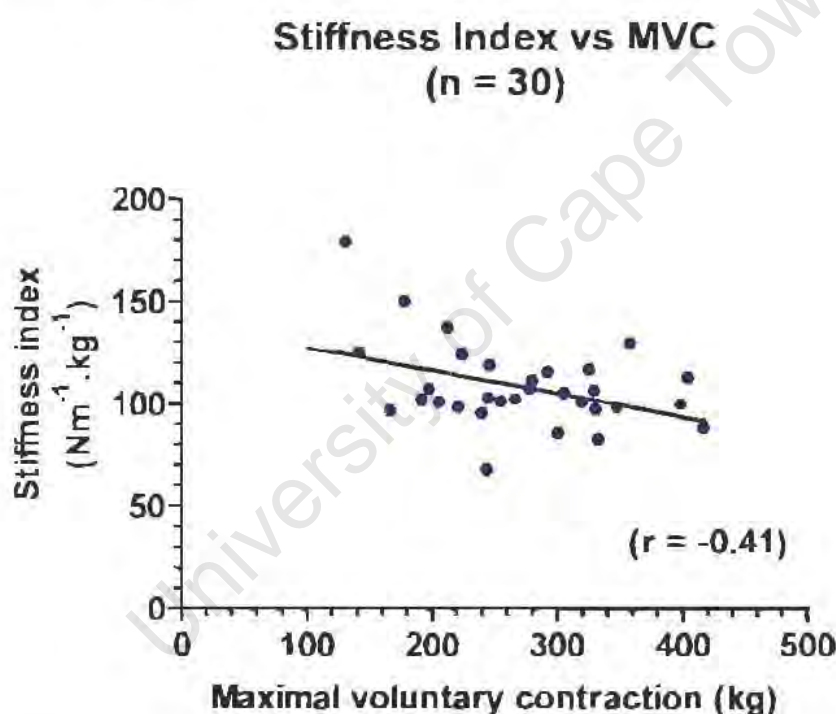


Figure 7: Relationship between MVC (kg) and body mass (kg) (n = 30)



When comparing the stiff group versus the compliant group, it was evident that there was a trend towards the stiffer group ( $83.3 \pm 10.4$  kg) being heavier than the compliant group ( $76.9 \pm 12.2$  kg). This difference was however not statistically significant ( $p < 0.14$ ).

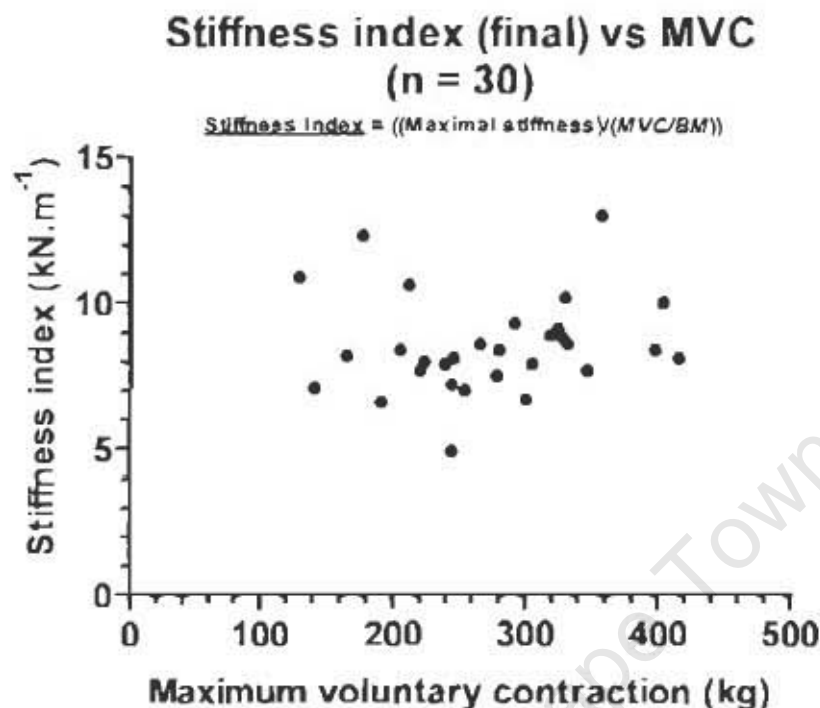
These findings suggest that both body mass and MVC need to be controlled before stiffness measurement can be compared between subjects. In accordance with this, we compensated for the magnitude of the load by creating an index or ratio, which would normalize the effect of MVC on stiffness. The maximal stiffness values ( $\text{N.m}^{-1}$ ) were divided by the respective individual's MVC (kg) to create the relative stiffness index ( $\text{N.m}^{-1}.\text{kg}^{-1}$ ). A weak, but significant relationship ( $r = -0.41$ ,  $p < 0.02$ ) was found between this relative stiffness index and the subjects' MVC (Figure 8). Only 17% of the variance in the stiffness index could be explained by the individual's MVC.



**Figure 8:** Relationship between stiffness index (normalized for load) and MVC (n = 30).

In an attempt to reduce the effect of MVC on stiffness even more, another normalized stiffness index was derived, by dividing the absolute maximal musculotendinous stiffness ( $\text{kN.m}^{-1}$ ) by a unit-less strength/mass ratio (MVC (kg)/body mass (kg)). There was no relationship between this normalized stiffness index and MVC (Figure 9) ( $r = 0.10$ ,  $p < 0.59$ ). It is clear from Figure 9 that the relationship between stiffness and MVC is

negated, by normalizing maximal stiffness for the strength/mass ratio. This was further confirmed, as the normalised stiffness index of the stiff group ( $9.06 \pm 1.42 \text{ kN.m}^{-1}$ ) was not different from the compliant group ( $8.01 \pm 1.83 \text{ kN.m}^{-1}$ ) ( $p < 0.09$ ).



**Figure 9:** Relationship between stiffness index ( $\text{kN.m}^{-1}$ ) and MVC (kg). The stiffness Index was calculated by dividing maximal stiffness ( $\text{kN.m}^{-1}$ ) by the strength/mass ratio (MVC (kg)/body mass (kg)).

## DISCUSSION

This first main finding of this study was that the calculation of the elastic stiffness of the musculotendinous complex and tendons of the lower body musculature was repeatable. This conclusion can be made based on the high intra-class correlation coefficients for the stiffness and low coefficients of variation between the various measurements conducted on ten subjects on 3 different occasions (Table 3). The finding for maximal stiffness ( $R = 0.97$  and  $CV = 4.9$ ) was similar to the reliability determined using a similar piece of equipment (151). Walshe et al. (151) found an intra-class correlation coefficient of  $R = 0.94$  and a coefficient of variation of 8% on their apparatus. The average stiffness in this study ( $28.3 \pm 7.3 \text{ kN.m}^{-1}$ , ranging from 16.0 - 46.0  $\text{kN.m}^{-1}$ ) was higher than that recorded by Walshe et al. (151) who had a mean of  $16.2 \pm 4.9 \text{ kN.m}^{-1}$ . However, these

values are within the typical reported ranges (10 - 50 kN.m<sup>-1</sup>) of stiffness for the combined lower body musculature (32;65;156).

Even though the maximal stiffness data has been shown to be quite repeatable, one should also take note of the range of precision of measurement for future testing. Determined by the criteria of Bland-Altman (16), the precision of measurement of maximal stiffness of the lower body, falls within a range of 10.6 kN.m<sup>-1</sup>. So for any differences to be measured, this needs to be taken into consideration (Figure 1).

The range of damped natural frequency data lies between 1.79 and 2.06 Hz (Table 6). These data compare well with the frequency data of the upper body musculature of Wilson et al. (156). They also compare with the data of Greene and McMahon (65), Shorten (132) and Cavagna (32) of the lower body. The induced oscillations in the study of Greene and McMahon (65) ranged between 1 and 3 Hz using a loaded position similar to the mid-stance phase during running. Cavagna (32) measured the elastic stiffness of the calf musculature during landing and showed that oscillation frequency varied from  $2.5 \pm 0.1$  to  $3.6 \pm 0.1$  Hz. Shorten (132), also using the calf musculature and isolating the soleus muscle in a bent-knee position, recorded frequencies ranging from 3 - 6 Hz. Wilson et al. (156) recorded oscillation frequencies ranging from 1.8 to 2.4 Hz using the upper body musculature in a loaded bench press position and using a variety of loads.

The second finding of this study was that as the load increased, the oscillation frequency decreased (Table 6). This finding is supported by other studies. For example, Wilson et al. (156) showed a similar tendency in their study. Their oscillation frequency data started at  $2.3 \pm 0.3$  Hz at 15% of maximum load and decreased to  $2.0 \pm 0.2$  Hz at 70% of maximum load. Cavagna (32), with two-legged landings registered  $3.6 \pm 0.1$  Hz, with one-legged landings registered  $2.9 \pm 0.2$  Hz and with loaded one-legged landings registered  $2.5 \pm 0.1$  Hz. Here the same pattern can be seen, where the relative load dispersed over two legs is less than over one leg, and the loaded one-legged landing is relatively heavier than the unloaded one-legged landings. If the oscillation frequency remained the same or increased with increasing loads on the muscle-tendon complex, then stiffness would continue to increase until infinity. This makes sense, as the oscillation frequency is one of the most important determining factors in the calculation of muscle-tendon complex stiffness. When one examines the formulae (b) through to (e) (p 133), this is evident.

The mass or load on the system also has an impact on the calculation of stiffness. However, the curvilinear representation of the stiffness-load relationship of the series elastic elements of muscle and tendon (Figure 2) shows that this relationship reaches a plateau. This plateau has been confirmed in previous research regarding the stiffness-load curve of muscle and tendon (132;153;156). Musculotendinous stiffness increases to maintain increasing load as the muscles are loaded with heavier weights (115). Eventually the stiffness of the muscle reaches that of the tendon, which becomes constant at higher forces (38) and is more compliant than active muscle (124;137). As this constant stiffness is approached, i.e. stiffness of the tendons, a plateau begins to form in the relationship between stiffness and load. This plateau at higher forces then represents the stiffness of the more compliant tendons, rather than the active muscles (132). The factor that relates to this gradual formation of a plateau in stiffness seems to be the increased time-periods between oscillatory force-peaks as the loads are increased, and this is reflected in the related decrease in damped natural frequency (Figure 3). The balance between the increase in load and decrease in damped oscillation frequency determines the slope of the exponential curvilinear association and plateau formation.

The third important finding in this study was the positive relationship between the strength of the individual and maximal musculotendinous stiffness (Figures 4 and 5). As the oscillation technique in its present format involves relative loads based on the strength of the subject, this implies that the higher the subject's MVC, the higher the absolute loads on the system and therefore also the higher the relative and maximal stiffness values. This finding raises questions about the interpretations of data from previously published research using this technique. For example, when stiff versus more compliant individuals are compared, it results in a comparison between strong versus weak individuals (Figure 6). Therefore it seems that the previous studies using this technique (149;151) may have reached the wrong conclusions because they were simply comparing strong vs. weaker subjects.

Furthermore, based on the significant relationship between MVC of the lower body musculature, and body mass of the subject (Figure 7), one has to question the interpretation of these studies. On the basis of resistance training, a heavier subject will load the muscles of the lower limbs more than a subject who does not weigh as much, resulting in stronger legs. If one examines the groups from the study of Walshe and

Wilson (149) for stiff versus compliant individuals, the stiff subjects weigh  $77.9 \pm 7.4$  kg and the compliant subjects  $71.9 \pm 6.5$  kg. As there is no mention about the group MVC averages, one has to assume on the basis of the findings in this study (Figure 7), that the stiff group was considerably stronger than the compliant group.

Although the technique of measuring musculotendinous stiffness has been confirmed through previous studies (32;132;149;151;153;156), the interpretation of some of their data are questionable. It follows that musculotendinous stiffness is higher at heavier loads, as the tendons of stronger individuals need to be stiffer to maintain the forces applied against them, and transfer these forces generated by the muscles to the bones. This however raises a question on the feasibility of measuring musculotendinous stiffness in the way described without normalizing the data for the subject's body mass or MVC. Therefore, the relative stiffness index, which measures relative stiffness by negating the effects of body mass and muscle strength as shown in Figure 9, offers a viable method of normalizing the musculotendinous stiffness values so that comparisons can be made between subjects who may differ in body mass and strength.

An example of the possible misinterpretation of stiffness data is shown in the study of Walshe and Wilson (149). In this study they compared the effects of musculotendinous stiffness, i.e. stiff versus compliant, on the stretch shortening cycle ability of individuals. They used a measure of relative drop jump height as an indicator of stretch shortening cycle ability. They compared drop jumps from various heights, i.e. 20, 40, 60, 80 and 100 cm, and normalized the results as a percentage of countermovement jump height. The stiff group gradually decreased more than the compliant group in the relative drop-jump heights, as the drop height increased. The difference between the two groups only became significant at drop heights of 80 and 100 cm.

Their interpretation was that the stiffer individuals had a marked deterioration in their toleration of high loads and that this finding was partly explained by the neural protective mechanisms of the Golgi-tendon organ. Based on the reduction of force output in the stiff individuals, they further suggest that the underlying stiffness might have significant implications in the expression of force. All these results may indeed be correct, but one factor that remains unclear is the relationship between stiffness and MVC, and MVC and body mass. As mentioned earlier, their stiffer group ( $77.9 \pm 7.4$  kg) was heavier than their



compliant group ( $71.9 \pm 6.5$  kg). On the basis of the findings reported earlier, the stiff group was most probably stronger (Figure 4, 5 and 6) and heavier (Figure 7). Hence the finding in their study might be reduced to the effect of body mass rather than of stiffness.

The impact force of a heavier individual would increase substantially more than a lighter individual as the drop height increases. Hence the rationale of the Golgi-tendon organ protective mechanism might still be correct, only for different reasons. As the Golgi-tendon organ is sensitive to the force applied on the muscle and tendon i.e. force of stretch, this might indeed be true. On the basis of the different interpretations of our study, we question the earlier findings of Walshe and Wilson (149;151).

The index of stiffness that was developed in this study has the potential to compare subjects of different strengths and body masses (Figure 9). This index could possibly categorize the relative stiffness of subjects and could contribute to more mechanistic studies on factors influencing musculotendinous properties.

Further research is however needed to determine if this indeed provides a way of comparing musculotendinous stiffness in subjects of different mass and strength.

## **CONCLUSION**

The oscillation technique as a measure of musculotendinous stiffness of the lower body extremities using the NAMS unit (Zest manufacturing, PTY (Ltd) and University of Cape Town, South Africa) was found to be repeatable. Absolute musculotendinous stiffness in its present format separates strong from weak individuals, instead of relatively stiff from relatively compliant individuals. Therefore, the practicality and usefulness of this measure, as it has been used in other studies is debatable, as it has led to various misinterpretations on the effect of musculotendinous stiffness especially on the stretch shortening cycle ability of muscle. It is recommended that musculotendinous stiffness be corrected for load and body mass for relative comparisons between subjects. Further research is needed to determine if an index of stiffness can indeed compare the relative elastic stiffness amongst individuals, and if it can be used to predict individual stretch shortening cycle ability.

# THE UPPER BODY

## METHODS

### SUBJECTS

13 Male subjects of various training backgrounds were used for this study. Of the 13 subjects, one subject could not master the oscillation technique, one subject dropped out of the study and two subjects could not bench press the minimum load of 70 kg required for this testing procedure. The minimum load was defined by the mass of the bar-tracking unit (21.32 kg). The mass of 30% of maximum load of 21.32 kg equates to a maximum (1RM) rebound bench press load of  $\pm 71$  kg. Hence these individuals were excluded from the study.

The remainder of the group ( $n = 9$ ) ranged between the ages of 20 and 32 years. Their personal characteristics are summarized in Table 9. Body fat was measured and expressed as a % (43) and as a sum of 7 skinfolds (129).

**Table 9:** Personal characteristics of subject group ( $n = 9$ )

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)	RBP (kg) (*)
26.3 $\pm 3.7$	179.4 $\pm 7.9$	83.0 $\pm 8.6$	74.7 $\pm 22.5$	15.3 $\pm 4.0$	70.1 $\pm 6.8$	93.5 $\pm 13.9$

(\*) Rebound Bench Press

Subjects were required to have resistance training experience, i.e. have trained in a gymnasium for at least one year. The subjects were tested on four different occasions.

The first day was a familiarization day, in which the complete testing procedure was performed as a trial run. This was performed, as described in the lower body test, to ensure that the subject knew completely what was expected of him, as well as to enhance the reliability of the testing procedure. The rebound bench press (RBP) test was also performed on this first day to determine the loads used in the familiarization trial as well as the days that followed. After the familiarization test, the subjects had to visit the laboratory on a further three occasions where the oscillation test was repeated.



To ensure that the subjects had fully recovered from the previous day's testing, a 48-hour break was given between testing days. Subjects were asked to avoid any upper body training at least 24 hours before testing and to refrain from any heavy or intense training throughout the duration of the study. All subjects were informed as to the nature of the study and the potential risks involved. The Ethics Committee of the University of Cape Town approved the study. Each subject completed a personal questionnaire regarding medical, training and general information and further signed a written informed consent form complying with the guidelines set by the American College of Sports Medicine, before participating in the study (3).

## **EXPERIMENTAL DESIGN**

On the first day body composition and maximal rebound bench press were measured. After the initial tests were performed, the subjects were given a familiarization session. During this session the various oscillation technique procedures were demonstrated. The subjects performed between 3 and 6 trials per load to perfect the technique and become familiar with the procedure and what was expected from them.

On the second testing day, the subjects were tested for maximal musculotendinous stiffness of the upper body. On the third and fourth day, the same testing procedure for maximal musculotendinous stiffness was used as for the second day. Each day's testing was preceded by a standardized warm-up. The standardized generalized warm-up consisted of 5 minutes of self-paced, submaximal shuttle-runs, 2 sets of 10 repetition standard push-ups, specific stretches for the upper body and 1 set of 10 repetitions bench press with the unloaded bar (21.32 kg) of the NAMS unit.

### **TESTING FOR THE REBOUND BENCH PRESS (RBP)**

For the upper body, the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town, Cape Town, South Africa) was set up and the subject positioned as previously described in Chapter 2 (p 118 - 119). This position was used to simulate a functional measure of strength using muscle groups in a similar position to that used during pressing movements.

Each subject was given 1 - 2 submaximal warm-up trials, where they were instructed to push a load of at least 80-90% of their predicted maximum. Two to five trials per subject followed the warm-up, with 2 - 3 minutes rest between trials. The mass on the bar was incrementally increased within 2 - 5 trials until the subject could no longer lift the bar to completion (130). The subjects were instructed to quickly lower the bar and on hearing the sound trigger, reverse direction and push the bar out until the arms were locked above the chest. The same exclusion criteria were used as described in Chapter 2 (p 120). The maximum load lifted successfully was recorded and used in the measurement of upper body stiffness.

### **THE OSCILLATION TECHNIQUE**

For the upper body oscillation test the procedure described in Chapter 2 (p 118 - 120) was used and subjects were instructed to maintain a constant muscular activity and force and not to respond to the perturbation in any way (106;156).

The force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

The same procedures and exclusion criteria as used for the lower body were implemented in the calculation of musculotendinous (MTC) stiffness of the upper body.

### **ELECTROMYOGRAPHY (EMG)**

EMG was also used in this technique to negate the effect of neurological intervention or contribution in the stiffness measures (149;151;155;156). Surface EMG was therefore measured before and after the perturbation, using LabVIEW 6.0.2 (National Instruments, Austin, Texas, USA) software and a telemetric EMG system (Noraxon, USA, Inc), to identify trials, which were confounded by neurological intervention (149;151;155;156).

The anterior deltoid was used as a control measure for neural intervention. Two surface EMG electrodes (blue sensor SP-00-S, Medicotest A/S, Rugmarken, Denmark) were attached as a pair over the muscle belly with an inter-electrode distance of ~2 cm (135) and were placed so that they were orientated parallel with the active muscle fibers. A third

neutral reference electrode was placed on the clavicle. Electromechanical delay was treated as a systematic error, as it was assumed that this was either negligible or constant (60).

The site for electrode placement was determined as the center of the muscle belly when contracted isometrically. The subject was prepared using the same procedure described for the lower body musculature (p 135). The same exclusion criteria were also used.

This oscillation test procedure was repeated the same way on all of the testing days.

## STATISTICAL ANALYSIS

The data were analysed using the same statistical procedures outlined for the lower body (p 135 - 137).

## RESULTS

Table 10 shows the average stiffness measure for all three trials at all the relative loads for each individual.

**Table 10:** Average stiffness data ( $\text{kN.m}^{-1}$ ) at each relative load for each individual between three trials ( $n = 9$ ). CV100% represents the coefficient of variation of maximal stiffness. Data were expressed as mean  $\pm$  SD.

SUBJECT	30%	40%	50%	60%	70%	100%	CV100%
Upper 1	5.7 $\pm$ 0.3	7.1 $\pm$ 0.2	8.3 $\pm$ 0.3	9.4 $\pm$ 0.5	10.4 $\pm$ 0.8	12.6 $\pm$ 1.9	14.8
Upper 2	6.7 $\pm$ 0.3	8.5 $\pm$ 0.2	10.1 $\pm$ 0.1	11.5 $\pm$ 0.4	12.8 $\pm$ 0.7	15.9 $\pm$ 1.8	11.3
Upper 3	8.0 $\pm$ 0.9	9.9 $\pm$ 1.0	11.4 $\pm$ 1.2	12.8 $\pm$ 1.3	13.9 $\pm$ 1.4	16.3 $\pm$ 1.9	11.5
Upper 4	7.4 $\pm$ 0.6	8.6 $\pm$ 0.8	9.5 $\pm$ 1.0	10.2 $\pm$ 1.2	10.7 $\pm$ 1.4	11.5 $\pm$ 1.7	15.0
Upper 5	8.0 $\pm$ 0.5	9.7 $\pm$ 0.4	11.1 $\pm$ 0.3	12.3 $\pm$ 0.3	13.2 $\pm$ 0.5	15.1 $\pm$ 1.2	7.8
Upper 6	8.2 $\pm$ 0.8	9.8 $\pm$ 0.8	11.1 $\pm$ 0.8	12.2 $\pm$ 0.8	13.0 $\pm$ 0.8	14.6 $\pm$ 0.7	5.0
Upper 7	5.9 $\pm$ 0.2	7.1 $\pm$ 0.2	8.0 $\pm$ 0.1	8.8 $\pm$ 0.1	9.4 $\pm$ 0.2	10.5 $\pm$ 0.5	4.6
Upper 9	7.7 $\pm$ 0.1	9.3 $\pm$ 0.3	10.5 $\pm$ 0.6	11.5 $\pm$ 1.0	12.3 $\pm$ 1.4	13.8 $\pm$ 2.4	17.1
Upper 10	5.0 $\pm$ 0.1	6.6 $\pm$ 0.1	8.2 $\pm$ 0.1	9.7 $\pm$ 0.0	11.2 $\pm$ 0.1	15.5 $\pm$ 0.7	4.2
Mean	7.0 $\pm$ 1.2	8.5 $\pm$ 1.3	9.8 $\pm$ 1.4	10.9 $\pm$ 1.4	11.9 $\pm$ 1.5	14.0 $\pm$ 2.0	10 $\pm$ 4.9

The intra-class correlation coefficient and the average coefficient of variation for maximal stiffness was  $R = 0.81$  (95% CI: 0.53-0.95) and  $CV = 10.1$  (95% CI: 7.5-15.4)% respectively (Table 11).

**Table 11:** Repeatability measures of musculotendinous stiffness of the upper extremities ( $n = 9$ ).

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
MTC stiffness@30%RBP	0.94 (0.83-0.98)	5.7 (4.3-8.7)
MTC stiffness@40%RBP	0.94 (0.83-0.98)	4.8 (3.6-7.3)
MTC stiffness@50%RBP	0.93 (0.80-0.98)	4.7 (3.5-7.2)
MTC stiffness@60%RBP	0.91 (0.75-0.98)	5.5 (4.1-8.4)
MTC stiffness@70%RBP	0.88 (0.68-0.97)	6.7 (5.0-10.2)
MTC stiffness@100%RBP	0.81 (0.53-0.95)	10.1 (7.5-15.4)

**Index:** MTC = Muscle-tendon complex, RBP = rebound bench press

In the upper body musculotendinous stiffness test, there were no significant differences between the various measures during the trials on different days (Table 12).

**Table 12:** One-way analysis of variance in the upper body musculotendinous stiffness test ( $n = 9$ ).

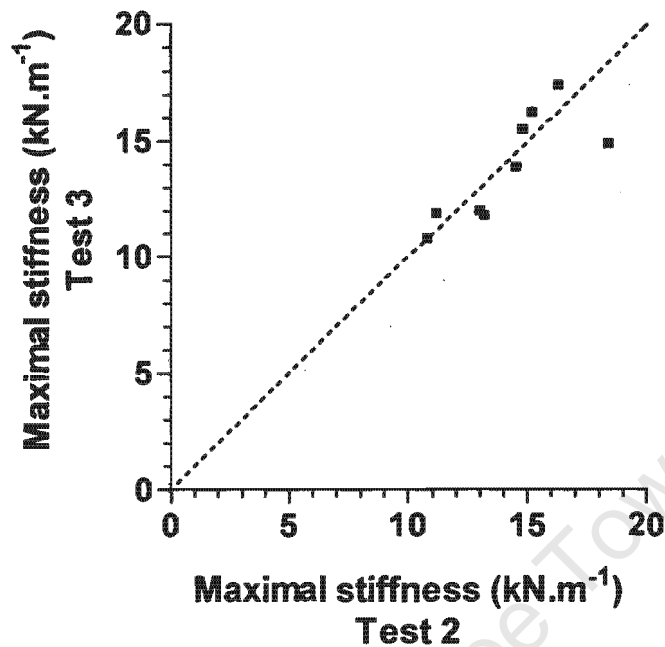
Variable	df effect	MS effect	df error	MS error	F	p-level (*significant)
MTC @30%RBP	2	0.22	16	0.25	0.87	0.44
MTC @40%RBP	2	0.31	16	0.29	1.07	0.37
MTC @50%RBP	2	0.38	16	0.39	0.98	0.40
MTC @60%RBP	2	0.41	16	0.58	0.70	0.51
MTC @70%RBP	2	0.39	16	0.91	0.43	0.66
MTC @100%RBP	2	0.24	16	2.63	0.09	0.91

**Index:** MTC = Muscle-tendon complex, RBP = rebound bench press

Figure 10 shows that the precision (two standard deviations above and below the average difference between trials) by limits of agreement, according to Bland-Altman (16), for measurement of maximal musculotendinous stiffness of the upper body falls within a range of  $6 \text{ kN.m}^{-1}$  ( $-0.3 \pm 3.0 \text{ kN.m}^{-1}$ ).

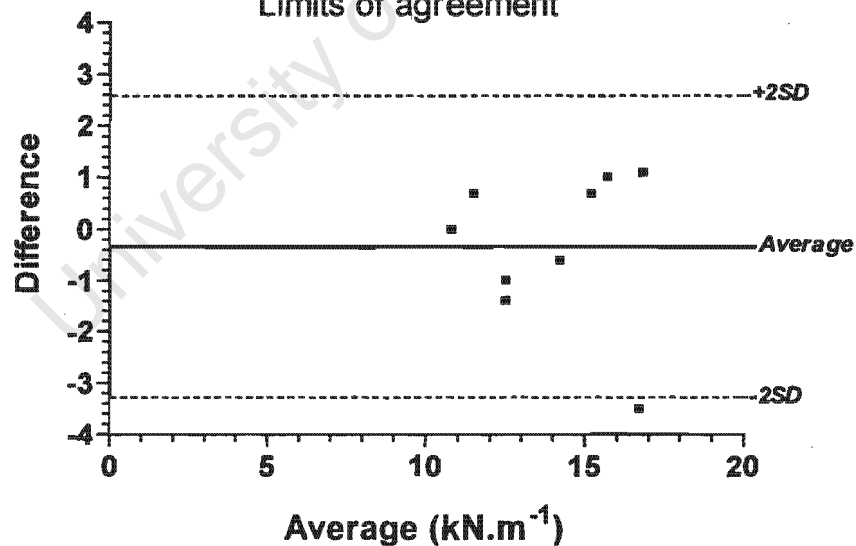
**A**

**Equality of measurement of  
maximal stiffness of the  
upper body**



**B**

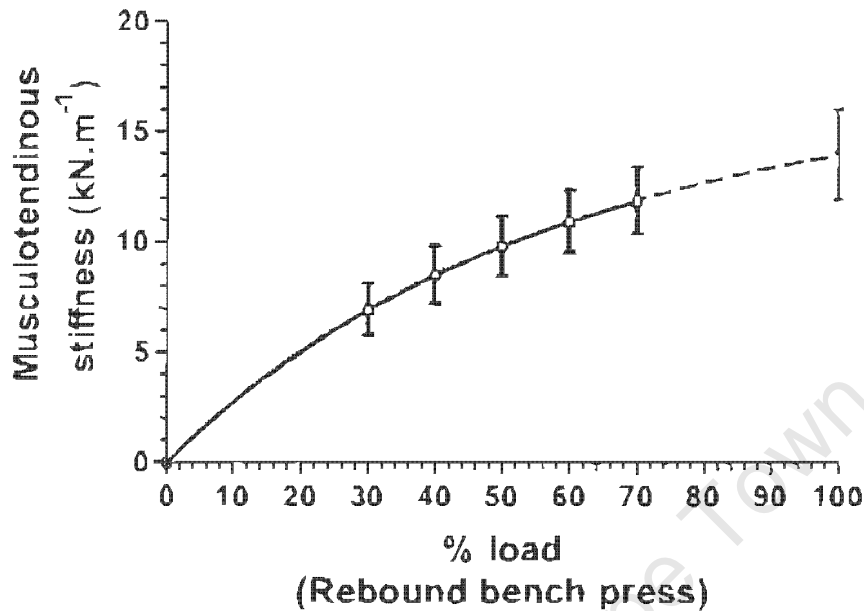
LOA (Maximal stiffness upper body) =  $-0.3 \pm 3.0 \text{ kN.m}^{-1}$   
Limits of agreement



**Figure 10:** A - The equality of measurement for maximal stiffness between the second and third testing day using the upper body oscillation test. B – The limits of agreement for measurement of maximal stiffness between the second and third testing day using the method of Bland-Altman (16).

As with the lower body results, the relationship between %MVC (defined by the rebound bench press) and musculotendinous stiffness was curvilinear (Figure 11).

### Group Profile (n = 9)



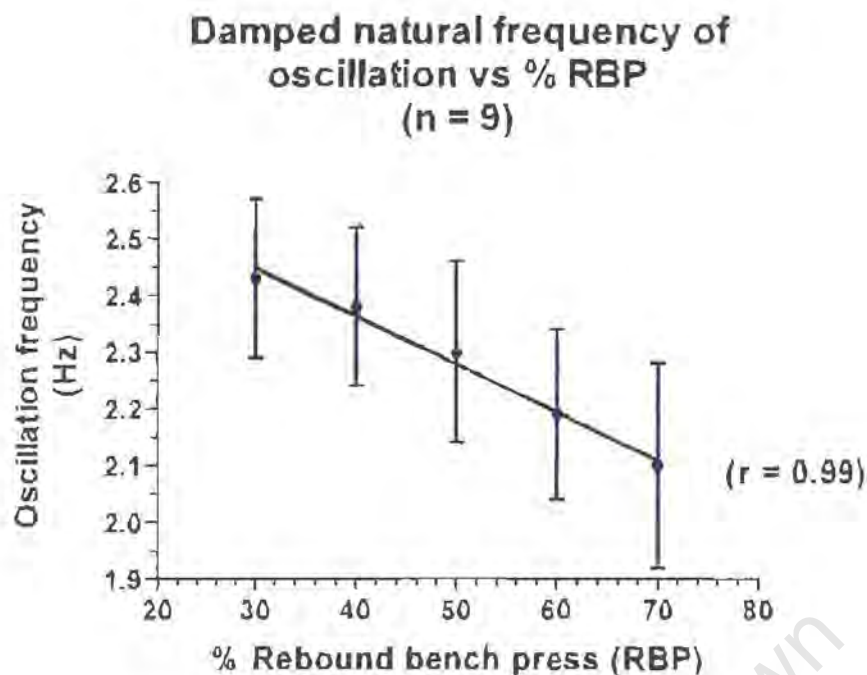
**Figure 11:** The curvilinear relationship between musculoskeletal stiffness and load (n = 9).

Furthermore, the frequency of oscillation decreased as the load on the muscle-tendon complex increased (Table 13 and Figure 12), as also shown by the lower body damped oscillation frequency data (Table 6).

The mean damped natural frequency of oscillation for the upper body stiffness test varied between 2.10 and 2.43 Hz over the various loading conditions (Table 13).

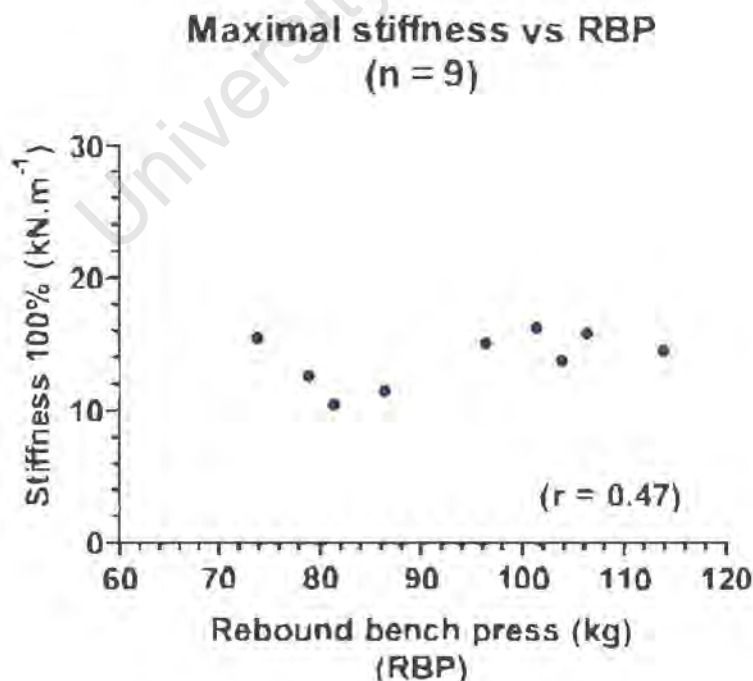
**Table 13:** The average damped natural frequency data at the relative loads on the NAMS unit, using the upper body maximal stiffness protocol (n = 9 x 6 trials).

% RBP	Damped Natural Frequency (Hz)
30%	2.43 ± 0.14
40%	2.38 ± 0.14
50%	2.30 ± 0.16
60%	2.19 ± 0.15
70%	2.10 ± 0.18



**Figure 12:** Relationship between % MVC, as defined by % rebound bench press (RBP), and oscillation frequency (Hz) ( $n = 9 \times 6$  trials per load).

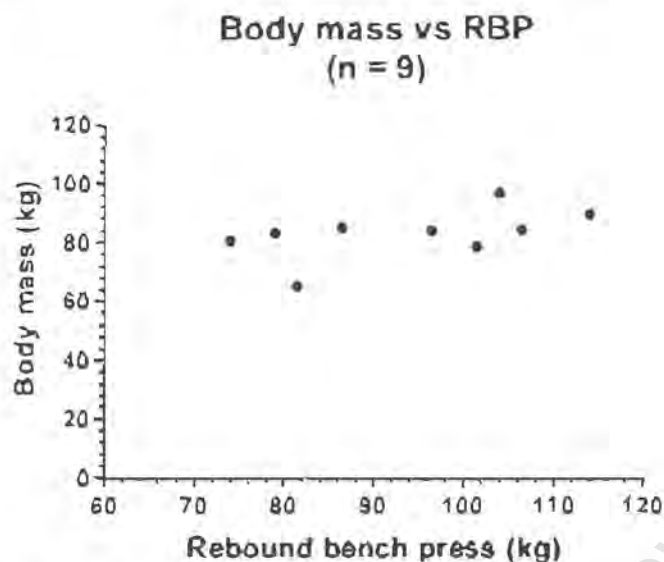
A weak, non-significant ( $r = 0.47$ ,  $p < 0.2$ ) relationship occurred between maximum voluntary contraction (defined by RBP) of the upper body and maximal musculotendinous stiffness (Figure 13).



**Figure 13:** Relationship between maximum voluntary contraction of the upper body defined by the rebound bench press (kg) and maximal MTC stiffness ( $n = 9$ ).



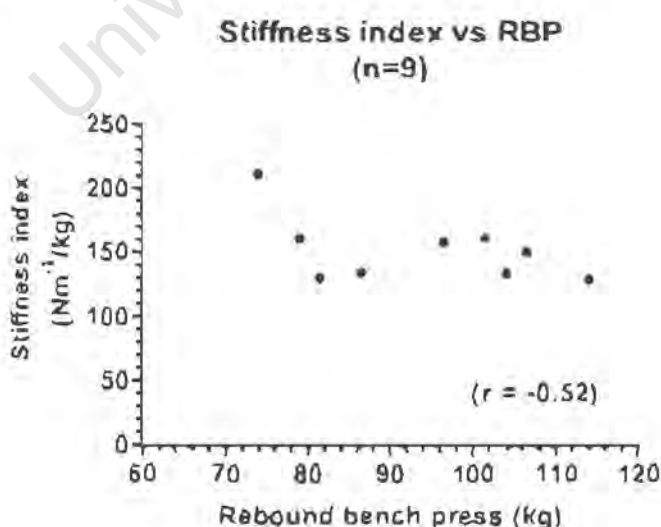
Body mass was also compared with maximal upper body strength, to determine if body mass had any relationship with upper body strength (Figure 14).



**Figure 14:** Relationship between maximum voluntary contraction of the upper body defined by the rebound bench press (kg) and body mass (kg) (n = 9).

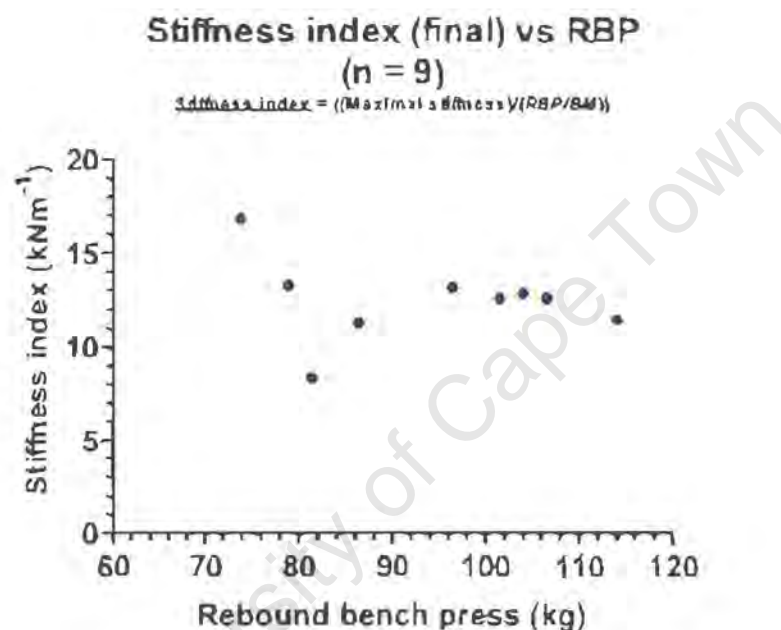
There was a weak ( $r = 0.54$ ), non-significant relationship ( $p < 0.13$ ) between the body mass of the subject and upper body strength (Figure 14).

Although there was no significant relationship between stiffness and load (Figure 13), we compensated for load as it does influence stiffness, however insignificant (refer to formulae (a) - (e), p 132 - 133). This initial stiffness index vs. rebound bench press is represented in Figure 15.



**Figure 15:** Relationship between the stiffness index (normalized for load) and maximum voluntary contraction of the upper body (n = 9).

As with the lower body test, this index did not seem to fit the criteria of compensating for the most important variables. There seemed to be a trend ( $p < 0.16$ ) between the load and stiffness index ( $\text{N.m}^{-1}.\text{kg}^{-1}$ ) so we attempted the second version of stiffness index ( $\text{kN.m}^{-1}$ ), by compensating for the relative strength of the individual. As for the lower body, the maximal strength value (rebound bench press) was divided by the body mass of the individual to obtain the ratio of strength relative to body mass. The maximal musculotendinous stiffness values were then normalised by dividing them by the relative strength ratio. No relationship ( $p < 0.59$ ) was found between this index and load (Figure 16).



**Figure 16:** Relationship between stiffness index ( $\text{kN.m}^{-1}$ ) and maximum voluntary contraction (kg) of the upper body. The stiffness index was calculated by dividing the maximal stiffness ( $\text{kN.m}^{-1}$ ) by the strength/mass ratio (RBP (kg) / body mass (kg)).

## DISCUSSION

The first finding of this study was that the stiffness values of the upper body (Tables 10 and 11) are more variable throughout the ranges of submaximal loads than in the lower body tests (Tables 2 and 3). Furthermore, the heavier the load, the more variable the measure of musculotendinous stiffness became (Table 11), which also supports the findings of Wilson et al. (153;156).

The maximal musculotendinous stiffness values ( $14.0 \pm 2.0 \text{ kN.m}^{-1}$ ) in this study were comparable with those values found in previous research using a similar testing protocol.

Wilson et al. (155) measured values of  $10.4 \pm 2.8 \text{ kN.m}^{-1}$ , and Wilson et al. (153) measured maximal stiffness values of  $18.3 \pm 3.5 \text{ kN.m}^{-1}$  in their experimental group, and  $16.5 \pm 5.0 \text{ kN.m}^{-1}$  in their control group in one of their earlier studies. Wilson et al. (156) in their initial study using a similar technique found values for maximal stiffness of  $17.9 \pm 4.6 \text{ kN.m}^{-1}$  in their group of subjects. Based on these results it seems that the maximal stiffness values of the upper body are comparable with those reported in the literature.

Even though the maximal stiffness data has been shown to be quite repeatable, one should also take note of the range of precision of measurement for future testing. Determined by the criteria of Bland-Altman (16), the precision of measurement of maximal stiffness of the upper body, falls within a range of  $6.0 \text{ kN.m}^{-1}$  (Figure 10).

The damped natural frequency data recorded (Table 13) in this study, which ranges from 2.10 to 2.43 Hz, also compares well with what is reported in the literature (p 147). The damped oscillation frequency also shows a similar tendency, as previously discussed (p 147 and Figure 3), to decrease as the load on the musculature increases (Figure 12). This supports the theory that the decrease in damped natural frequency, as the loading on the musculature increases, contributes to the plateau formation of the stiffness-load curve of the series elastic components.

The next finding of this study was that the relationship between maximal stiffness and load (rebound bench press) (Figure 13), and between maximal strength and body mass (Figure 14) are significantly weaker than those in the lower body. This can perhaps be explained as the upper and lower body muscles have different functions. The lower body muscles are essentially weight bearing and locomotive muscles. These muscles have to be relatively strong and are affected by, and adapt to, the weight that they have to transport/support.

The upper body muscles on the other hand are essentially non-weight-bearing and are associated more with a stabilization function, as well as powerful- and precision movements. Therefore, the weight-bearing lower body will have a stronger and more significant relationship with body mass than the non-weight-bearing upper body.

The relationship of load on maximal stiffness is however much less evident in the upper body. The weaker relationship between load and maximal stiffness in the upper body can also be explained by the homogeneity of the subject group that was tested. The range of maximal upper body strength amongst the nine individuals of this study (35%) was far lower than the group of the lower body study (69%). It is therefore possible that the spread of data is insufficient to determine a significant relationship. However, if one examines the formulae used (p 132 - 133) it is evident that the load plays an important role in the calculation of musculotendinous stiffness.

Wilson et al. (156) confirm this assumption as in their study they found a significant correlation ( $r = 0.71$ ,  $p < 0.05$ ) between maximum upper body strength, i.e. loads lifted (rebound bench press), and maximal stiffness of the series elastic elements. Their performance test data show that the stronger individuals are in the stiffer group and the weaker individuals in the compliant group (155). The stiff subjects had a concentric force output of  $981 \pm 83$  N and the compliant group  $872 \pm 180$  N. Keeping this in mind we initially normalized the stiffness data for load as we did for the lower body tests (Figure 15) and divided the maximal stiffness of the upper body by the maximum rebound bench press. However, this index was insufficient as a normalising factor, as there was a trend evident between rebound bench press and the stiffness index.

In an attempt to try and make the measurement of musculotendinous stiffness of the upper and lower body comparable, we calculated the stiffness index (Figure 16) in the same way it was calculated for the lower body data (Figure 9), i.e. the maximal stiffness was normalized for the relative strength of the individual. This index showed no significant relationship to the load for either the upper or the lower body. This suggests that the musculotendinous stiffness index meets the criteria to compare the relative maximal stiffness of the musculotendinous unit. As with the lower body, further research is necessary to determine if this stiffness index adequately controls for differences in body mass and strength.

## **CONCLUSION**

The oscillation technique on the upper body musculature using the NAMS unit (Zest manufacturing, PTY (Ltd) and University of Cape Town, South Africa) is a repeatable

measure of musculotendinous stiffness. Even though the findings of the present study show a lesser relationship between load and stiffness, compared to the lower body, when examining the literature of upper body musculotendinous stiffness it still seems that the interpretation of the data are flawed when the stronger subjects are compared to the weaker subjects.

## **SUMMARY OF FINDINGS**

The oscillation technique is a reliable measure of musculotendinous stiffness of both lower and upper body tests. However, the determination of absolute musculotendinous stiffness in its present format is influenced by the load against which the muscle is contracting, and the subject's body mass in the muscles of the lower body, and to a lesser extent in the muscles of the upper body. To measure and compare stronger with weaker individuals, an index of stiffness, which normalizes for the relative strength, might be a more sensitive and more valid measure of comparison of the elastic stiffness of individuals. Further research on this index of relative stiffness of the musculotendinous unit is needed to determine the efficacy and applicability of this measure.

## **CHAPTER 4:**

### **The modified oscillation technique using the NAMS unit**

#### **INTRODUCTION**

The oscillation technique used in the format described in Chapter 3 had numerous shortcomings that needed addressing. Although the measure of stiffness of the muscle-tendon complex using the NAMS unit (Zest Manufacturing, PTY (Ltd) and University of Cape Town, South Africa) was repeatable, four methodological concerns that needed attention, before the technique could be used in applied or mechanistic experiments, were identified after the experiment.

Firstly, not all subjects started to plateau in musculotendinous stiffness as the load increased. This can be explained by the fact that the loads were calculated as a percentage of maximum voluntary contraction and in certain subjects the maximum voluntary contraction was perhaps underestimated. Therefore, all their relative loads were lower than intended and did not load the muscle-tendon complex sufficiently to induce the formation of a plateau.

A second factor that could have influenced the results was that subjects wore shoes during the testing of musculotendinous stiffness. It was theorized that the cushioning of the shoes might absorb some of the load and influence the measurement of stiffness of the muscle-tendon complex in some way. Subjects initially wore shoes as a safety measure. However, with the experience of testing it was realized that the risk of injury was minimal, and that it was not necessary for subjects to wear shoes.

Thirdly, in the method described in Chapter 3, the stiffness values per load were plotted on a graph using the Graphpad Prism Version 3.0 software (Graphpad Software Inc., San Diego, California, USA) and the line of best fit was determined using a downward exponential association and least sum of squares criterion.

Maximal stiffness was further determined using the specific curve fit of the stiffness-load curve and the downward exponential association equation:

$$\text{Stiffness} = Y\text{-max} \times (1 - \text{Exp}^{(-K \times X)})$$

Where:

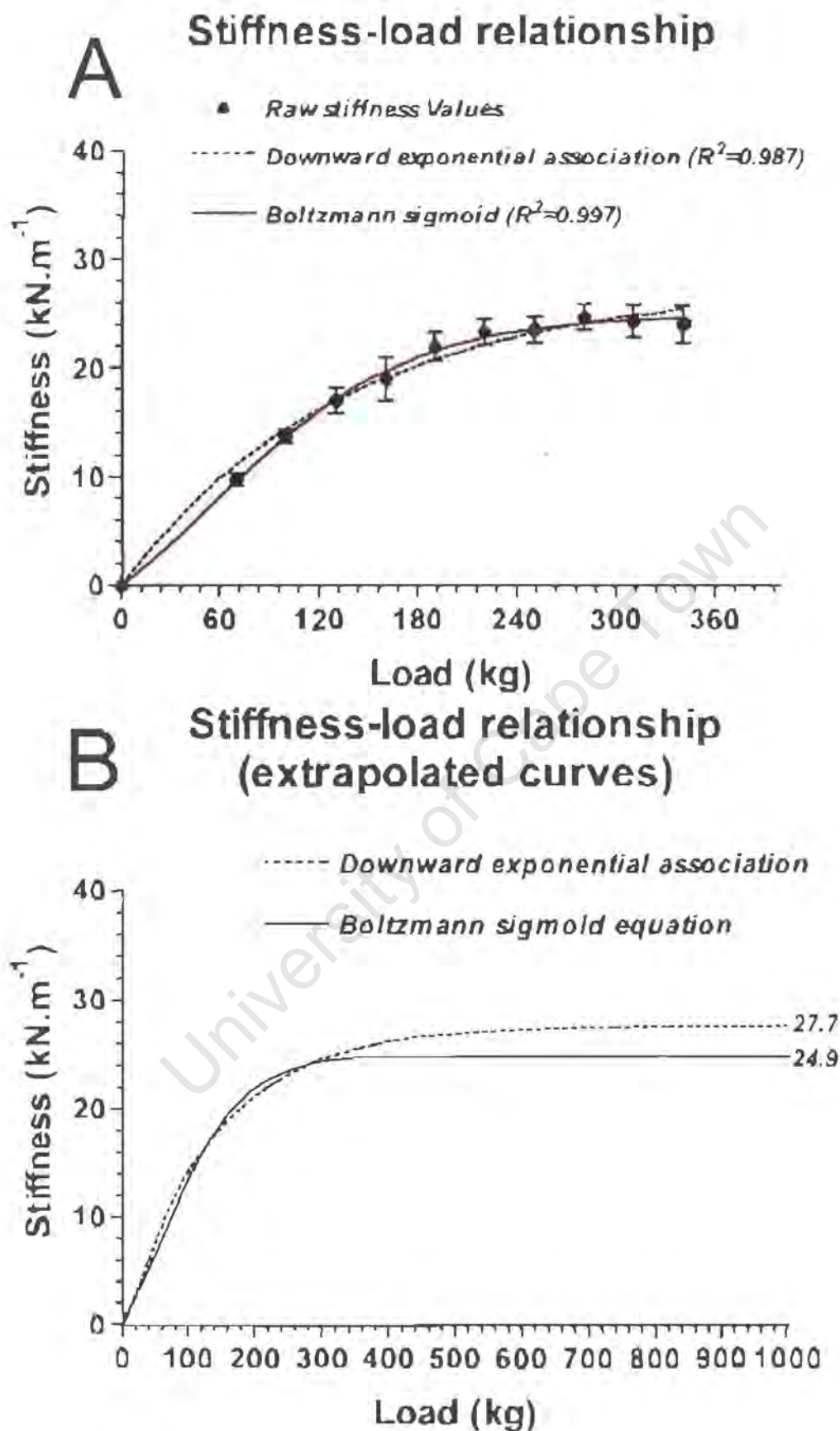
- **Y-max** = maximum plateau,
- **k** = a constant and
- **X** = percentage load.

According to this procedure, maximal stiffness was calculated from an extrapolated maximum load using the aforementioned equation. Even though the line of best fit was reasonably tight, this equation appeared to be rigid and forced the curve-fit to slightly diverge away from the true values at the upper range of loads. This resulted in a potential overestimation of the subjects' maximal stiffness or average stiffness of the tendons in the involved muscle groups.

After experimentation it was found that the use of the Boltzmann sigmoid equation in Graphpad Prizm Version 3.0 software (Graphpad Software Inc., San Diego, California, USA) described the relationship between load and musculetendinous stiffness better than the downward exponential association equation, which had been used in the previous experiment. Using the least sum of squares criterion and binding the curve to zero, the Boltzmann sigmoid equation fitted the data better than the downward exponential association equation. Therefore it was decided for future experiments to define the load vs. musculetendinous stiffness relationship with the Boltzmann sigmoid equation rather than the downward exponential association equation.

Figure 1 illustrates the differences between the two formulae when applied to the same raw data.





**Figure 1:** A - Shows the line of best fit using both the downward exponential association and Boltzmann sigmoid equations on the raw data of a theoretical subject. B - Shows the same curve fits of the identical data extrapolated to a maximum plateau.

In Figure 1A, it can be seen that the line of best fit of the Boltzmann equation is similar to the downward exponential association, and there is no major difference in the  $R^2$ -value ( $R^2 = 0.987$  vs.  $0.997$ ; downward exponential association vs. Boltzmann sigmoid). However, the Boltzmann sigmoid equation tracks the data better, with the line of best fit tracking more data points than the downward exponential association (Figure 1A).

In Figure 1A, it can also be seen that the line of best fit defined by the downward exponential association equation is rigid and seemingly overestimates the slope of both the initial portion of the stiffness-load curve as well as the latter portion, where the raw stiffness values start levelling off at maximum values. This has a major influence on the resultant extrapolated plateau value as can be seen in both parts A and B of Figure 1, and hence the downward exponential association equation overestimates the maximal values. It is logical to assume that the replacement of this equation with the Boltzmann equation might reduce the error associated with over-predicting.

The fourth factor that was identified as a potential problem was that the use of relative loads prevented the comparison of the stiffness of different individuals at the various absolute loads. To better understand morphological and functional differences between individuals, it might be more practical to develop a protocol using absolute loads.

Therefore the oscillation test described in Chapter 3 was modified to address these four issues in an attempt to refine the technique. The aim of this study was to test a group of subjects using an adapted protocol to determine if these modifications to the oscillation test were functional.

## **METHODS**

### **SUBJECTS**

14 Male subjects, between the ages of 22 and 32 years were recruited for this study. Although subjects had varying training backgrounds, they were required to have resistance training experience of at least one year. The Boltzmann sigmoid formula requires at least 5 data points to calculate the line of best fit and predict the maximum plateau or stiffness value. Two subjects could not achieve more than four loads and their data were therefore

excluded from further analysis. The general characteristics of the remaining subjects ( $n = 12$ ) are summarized in Table 1.

Body fat was measured and expressed as a % (43) and as a sum of 7 skinfolds (129).

**Table 1:** *Personal characteristics of subject group ( $n = 12$ )*

<b>AGE (yrs)</b>	<b>HEIGHT (cm)</b>	<b>MASS (kg)</b>	<b>SUM OF SKINFOLDS (mm)</b>	<b>% BODY FAT</b>	<b>LEAN BODY MASS (kg)</b>
25.7 $\pm 3.3$	180.2 $\pm 8.5$	80.1 $\pm 12.0$	62.4 $\pm 31.0$	14.0 $\pm 6.6$	68.8 $\pm 10.9$

All subjects were informed about the study and the potential risks involved. The Ethics Committee of the University of Cape Town approved the study. Each subject completed a personal questionnaire regarding medical, training and general information and further signed a written informed consent form complying with the guidelines set by the American College of Sports Medicine, before participating in the study (3).

### **EXPERIMENTAL DESIGN**

The first day was a familiarization day, in which the complete testing procedure was performed as a trial run. The subjects were given 4 - 6 trials per load to familiarize themselves with the procedure. The loads were increased incrementally to a load just less than their expected maximum tolerable load. This was performed to ensure that the subject knew what was expected of him, and to enhance the reliability of the testing procedure. After the familiarization test, the subjects had to visit the laboratory on one occasion where the test was repeated.

A 48-hour break was given between the familiarization day and subsequent testing days to ensure that the subjects had fully recovered. Subjects were asked to refrain from any lower body training at least 24 hours before testing and to refrain from any heavy or intense training throughout the duration of the study.

On the second day the modified oscillation test for the lower body musculature was performed to completion.

### THE MODIFIED OSCILLATION TEST FOR THE LOWER BODY

The same set-up and procedure was used as for Chapter 3 and previously described in Chapter 2 (p 121 - 122). Subjects were instructed to maintain a constant muscular activity and force and not to respond to the perturbation in any way (106;156). The fluctuation or oscillation in force registration via the load cell was used to calculate the stiffness of the muscle-tendon complex. Surface EMG was also measured to eliminate those tests that had noticeable pre-activation as this neural intervention would influence the stiffness measures (149;156). The main difference was that in the modified protocol, the subjects were not tested at relative loads calculated as a percentage of their maximum voluntary contraction, but at the same absolute loads. The initial load used was 70 kg. The subjects were further tested at incremental loads increasing by 30 kg per load until they could no longer maintain the load with sufficient stability. The calculated stiffness at the maximum load, i.e. where the subject became unstable, was compared to the rest of the data. If this value was a clear outlier, the data for this load was excluded from the ensuing calculation of maximal stiffness. The decision to exclude this data point was confirmed by an impartial reviewer. The test on average required the subjects to be tested at approximately 5 - 6 loads. The subjects performed the test without shoes to exclude the possible effect of cushioning.

The force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

The resultant damped oscillations were modelled on the same damped mass-spring model, and stiffness at each load was calculated with the same formulae utilised in Chapter 3 (p 132 - 133).

Six measurements were recorded at each load and using the procedure outlined in Chapter 2 (p 120), four of these measurements were used and averaged to determine the respective stiffness at each load. A rest of 2 - 3 minutes was allowed between each load and  $\pm 1$  minute between trials. Additional rest was given if required. The stiffness values per load were then plotted on a graph using the Graphpad Prism Version 3.0 software (Graphpad Software Inc., San Diego, California, USA) and the line of best fit determined using the Boltzmann sigmoid equation and least sum of squares criterion.

Maximal stiffness was determined as the upper plateau defined by the specific curve fit of the stiffness-load curve, using the equation:

$$\text{Stiffness} = \text{bottom} + (\text{top} - \text{bottom}) / (1 + \exp^{((V50 - X) / \text{slope})}) \quad (f)$$

Where:

- **bottom** = bottom plateau of the sigmoid curve,
- **top** = upper plateau of the sigmoid curve,
- **V50** = the point where the stiffness value is halfway between the bottom and upper plateaus,
- **slope** = the steepness of the curve (the larger the value the shallower the curve) and
- **X** = the load used.

This procedure of extrapolation has been recommended because most subjects cannot maintain the loads with sufficient stability at higher loads (149;151;153;155;156). We however, have modified this principle to using absolute loads and extrapolating the curve fit using the data points provided at the various incremental loads, until a plateau was reached. Maximal stiffness calculated this way, as with the original technique, was assumed to be representative of the averaged stiffness of the tendons of the muscles of the lower body (149;151).

#### **ELECTROMYOGRAPHY (EMG)**

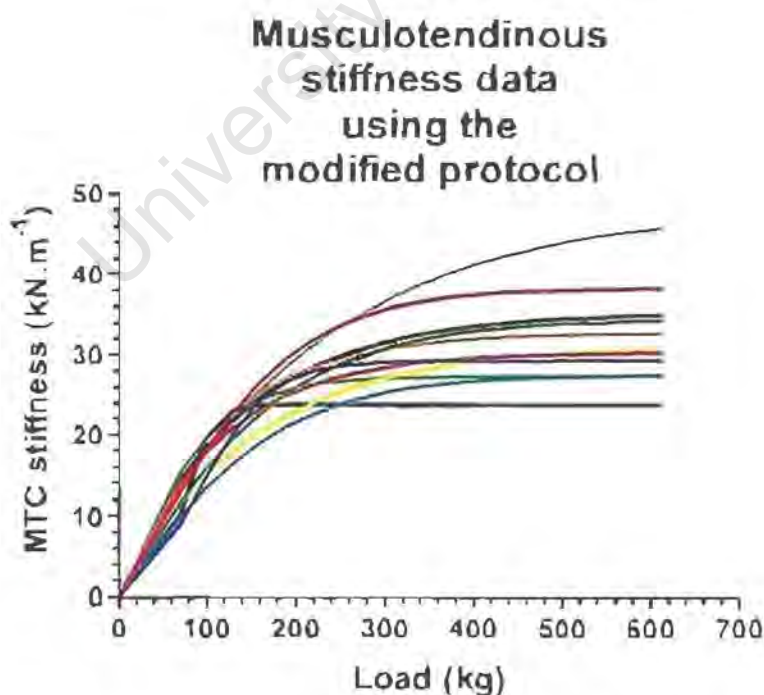
Surface EMG was again measured before and after the perturbation using LabVIEW 6.0.2 (National Instruments, Austin, Texas, USA) software and a telemetric EMG system (Noraxon, USA, Inc) to identify trials, which were confounded by neurological intervention (149;151;155;156). The same procedures were utilised as previously described in Chapter 3 (p 134 - 135).

## RESULTS

The modified oscillation technique presented a diverse range of maximal and submaximal stiffness measures amongst subjects. The data are tabulated in Table 2 and are graphically represented in Figure 2. Figure 2 also shows that predicted maximal stiffness does not seem to be related to its corresponding extrapolated maximal load.

**Table 2:** Musculotendinous stiffness values ( $\text{kN} \cdot \text{m}^{-1}$ ) of the subjects over the range of absolute loads and the maximal predicted stiffness of the subjects. ( $n = 12$ )

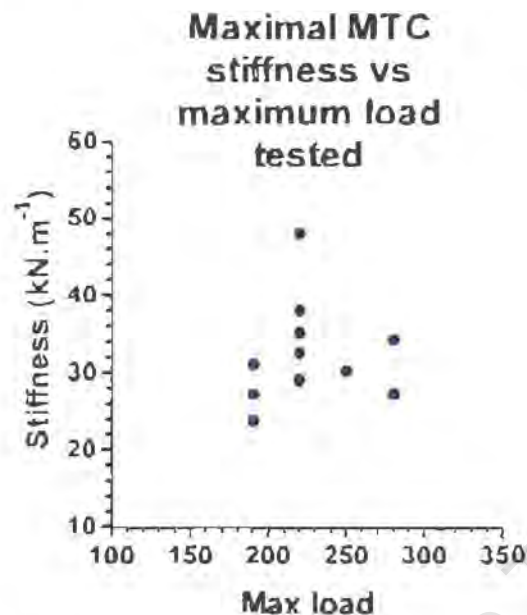
Subject	70kg	100kg	130kg	160kg	190kg	220kg	250kg	280kg	Maximal stiffness
Mod1	12.3	19.0	22.9	25.5	29.5	31.9	-	-	38.2
Mod3	13.4	18.3	21.4	22.0	23.7	26.2	-	-	29.2
Mod4	13.4	19.1	21.6	21.8	23.7	27.1	27.4	-	30.3
Mod5	11.0	16.1	20.9	22.4	23.5	27.3	29.7	29.7	34.4
Mod7	9.8	13.8	15.4	20.0	20.1	23.3	23.5	24.7	27.4
Mod8	12.0	15.7	17.4	18.6	23.7	-	-	-	31.2
Mod9	15.5	19.0	22.5	24.8	25.1	-	-	-	27.3
Mod10	10.2	19.8	22.6	23.6	24.2	-	-	-	23.8
Mod11	9.0	14.8	20.7	22.8	28.2	27.1	-	-	29.2
Mod12	16.7	18.9	22.1	25.0	26.4	29.1	-	-	32.7
Mod13	16.8	17.3	19.7	24.8	27.6	28.4	-	-	35.3
Mod14	13.5	19.0	20.0	25.3	29.3	30.8	-	-	48.1
Average	12.8	17.6	20.6	23.1	25.4	27.9	28.9	27.2	32.3
$\pm$ SD	$\pm 2.6$	$\pm 2.0$	$\pm 2.3$	$\pm 2.2$	$\pm 2.8$	$\pm 2.5$	$\pm 3.1$	$\pm 3.6$	$\pm 8.4$



**Figure 2:** The musculotendinous stiffness data of the subjects at absolute loads and extrapolated to their predicted maximal stiffness using the Boltzmann sigmoid equation ( $n = 12$ ).

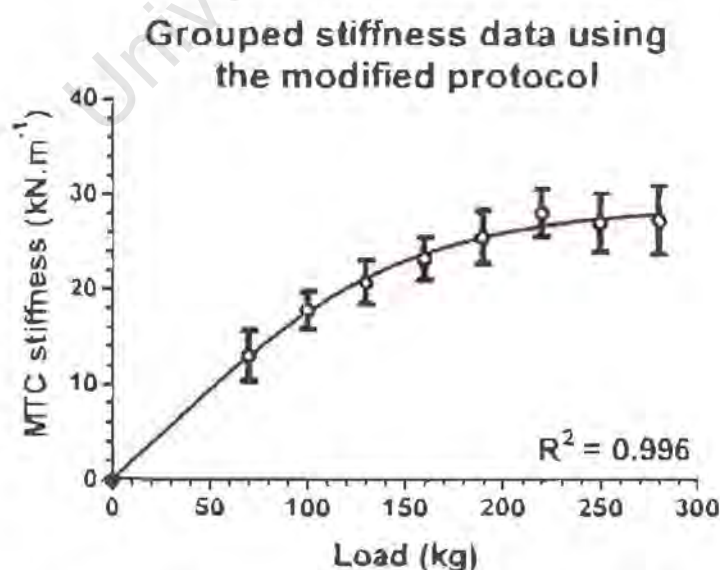


Figure 3 show the graph of maximal- or tendon stiffness and the maximum load obtained during the test and confirms that there was no relationship between these variables ( $r = 0.10$ ,  $p < 0.76$ ).



**Figure 3:** The relationship between maximal of tendon stiffness and the maximal load obtained ( $n = 12$ ).

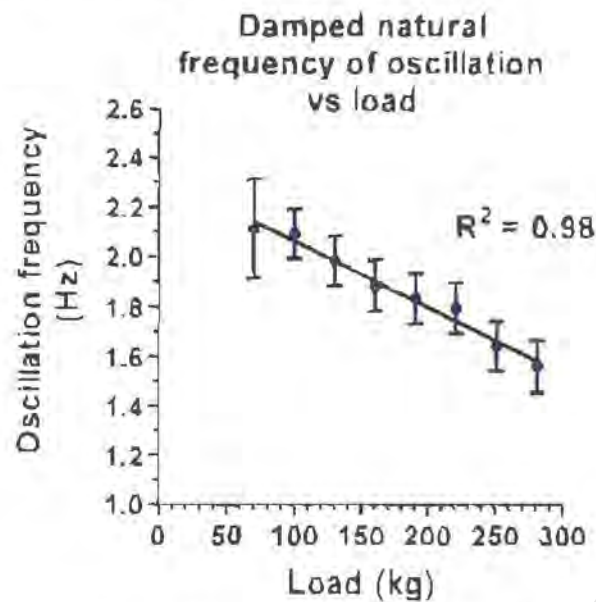
The general tendency of the stiffness-load curve of the series elastic elements of muscle and tendon (Figure 4) followed the classic curvilinear model, which was also found in earlier studies (151;153;156). The average maximal stiffness of the group was  $32.3 \pm 6.4 \text{ kN.m}^{-1}$ .



**Figure 4:** The averaged stiffness-load relationship of all the subjects used in this study ( $n = 12$ ).



Figure 5 and Table 3 show the inverse relationship between the damped natural frequency of oscillation and the load on the muscle-tendon complex.



**Figure 5:** The relationship between damped natural oscillation frequency and load ( $n = 12$ ).

Table 3 shows the damped natural frequency of oscillation ranged from 1.56 - 2.11 Hz.

**Table 3:** The average damped natural frequency data at the relative loads using the NAMS unit ( $n \times 6$  trials per load).

Absolute load (kg)	Damped Natural Frequency (Hz)
70	2.11 $\pm$ 0.20
100	2.09 $\pm$ 0.13
130	1.98 $\pm$ 0.12
160	1.88 $\pm$ 0.11
190	1.83 $\pm$ 0.12
220	1.79 $\pm$ 0.09
250	1.64 $\pm$ 0.10
280	1.56 $\pm$ 0.08

## DISCUSSION

The first major finding of this study is that the modified oscillation technique, using absolute rather than relative loads, gives a better representation of the plateau phenomenon (Figure 4), as shown in previous studies (132;153;156), than the original procedure used in Chapter 3. This can also be attributed to the change in the equation used to calculate the line of best fit (load vs. musculotendinous stiffness), from the downward exponential association to the Boltzmann sigmoid equation. The lines of best fit calculated from the two equations, did not differ significantly from each other in this subject group, i.e.  $R^2 = 0.99$  for the Boltzmann equation vs.  $R^2 = 0.98$  for the downward

exponential association. However, the shape of the Boltzmann graph followed a less rigid path and gave a more accurate representation of the functional properties expected of a muscle-tendon complex under loading conditions than the downward exponential association. This is particularly evident as demonstrated in Figure 1 and was also found using the subject data in this study.

The fact that the subjects performed the tests without shoes could also have contributed to this finding. It is possible that barefoot testing allowed the muscle-tendon complex to be more sensitive to the perturbation applied. It is further also possible, as the test was performed barefoot, that the results reflected the true elastic nature of muscle and tendon in the involved extremities, whereas previously the cushioning of the shoe absorbed some of the perturbation.

Additionally, by not depending on the maximum voluntary contraction test to estimate the relative loads, a sourced error, which accompanies this estimation of maximal strength, was avoided. By using absolute loads and incrementally testing subjects to their maximum sustainable loads, there were generally more data points than in the relative loading technique described in Chapter 3 to determine the line of best fit. This improved the accuracy of the measure. Also, by testing the subjects in this way any underestimation of maximal strength via the maximum voluntary contraction was avoided and the maximal load tolerated was accepted as being representative of the current training status of the subject.

The second finding of this study is that there are distinct differences between subjects in terms of their stiffness vs. load profiles (Table 2, Figure 2). This was not as clear using the relative loading technique described in our earlier studies (Chapter 3), as the loads were not the same. A comparison of the musculotendinous stiffness of the subjects at submaximal absolute workloads might provide additional information and contribute to a better understanding about the factors, which determine a muscle's contractile status or ability e.g. strength, power, stretch shortening cycle ability, susceptibility to muscle damage, inclination for injury, etc. The data also gives a significant spread of maximal- or tendon stiffness values between subjects.

Thirdly, there is no association between load and maximal/tendon stiffness using the modified technique as was found in the initial study with relative submaximal loads. An advantage of this technique is that the data do not have to be transformed using mathematical derivatives to calculate a stiffness index (Chapter 3). It may be concluded that the restrictions in the original downward exponential association formula (dominated by the initial linear portion of the curve), together with the use of relative loads, determined by an underestimated maximum voluntary contraction, might have contributed to this original conclusion. It would appear as if the use of absolute loads until muscle failure might be a more accurate measure of musculotendinous stiffness, and therefore also maximal musculotendinous- or tendon stiffness. However, based on the fact that two subjects were excluded on the basis that they could only attain four loads, it would be advisable to reduce the mass of the load increments for future studies.

The average maximal stiffness of the musculotendinous unit of  $32.3 \pm 6.3 \text{ kN.m}^{-1}$  are within the range of values ( $10 - 50 \text{ kN.m}^{-1}$ ) reported in the literature for maximal stiffness of the combined lower body musculature (32;65;156). The range of damped natural frequency values i.e. 1.56 - 2.11 Hz and the decrease in oscillation frequency with increased loading (Figure 5, Table 3) is consistent with that found in the literature (32;65;132;156).

## CONCLUSION

The modified oscillation technique using an absolute loading protocol described in this chapter seems to be a more accurate way of measuring musculotendinous stiffness than the original method, which used a relative loading protocol (Chapter 3, (149;151;153;155;156)).

Additional support for the use of the modified oscillation versus the original technique is that the Boltzmann sigmoid equation traces the data closer than the downward exponential association formula, and the absolute loads tend to generally give more data points for determining the line of best fit, which improves its accuracy and representation.

The maximum voluntary contraction procedure used in the original technique described in Chapter 3 underestimates the maximum strength in some subjects. This might contribute

to the muscle-tendon complex not being loaded sufficiently to induce the formation of a plateau. Using absolute loads and increasing the load until the subject cannot maintain it stable enough seems to have accounted for this error.

In conclusion it is therefore proposed that further studies using the oscillation technique for the upper and lower body musculature should follow an absolute loading- rather than a relative loading protocol.

University of Cape Town

## **CHAPTER 5:**

# **The repeatability of the stretch shortening cycle test using the NAMS unit in the lower and the upper body**

## **INTRODUCTION**

As mentioned in the literature review, Chapter 1, muscles are frequently exposed to impact or gravitational loading, which causes a cycling between eccentric and concentric contractions, where the concentric- is preceded by eccentric contraction (91;92;95). During most sporting activities there usually is a countermovement in the opposite direction (137) to prime the muscles before the movement is initiated in a specific direction.

The active muscles first have to absorb the momentum by eccentric contraction and then follow this immediately by a concentric contraction in the opposite intended direction (132). This cyclic relationship is referred to as the stretch shortening cycle (SSC), also known as plyometric muscle action (17;22) and is a natural component of muscle function (139).

This cyclic form of muscle action results in an increased muscle performance when compared to isolated concentric contractions without prior stretch (26;27;47;56;136-140). The eccentric contraction phase positively influences the subsequent concentric contraction resulting in a more powerful contraction (32;91;92;119;137;138). The higher force at the beginning of the concentric movement in stretch shortening cycle actions leads to an increased acceleration of the mass to which the muscle is attached and hence greater power output (33).

The most common technique for measuring stretch shortening cycle ability has been variations of vertical jumping techniques (13;73;89;95;135;149). These have ranged from the standard type of jumping tests, i.e. squat jumps (SJ); countermovement jumps (CMJ); and drop jumps (DJ) from various heights (7;13;21;24;26;62;73;89;144;149;151). Vertical jumping techniques utilise the stretch

shortening cycle as a mechanism for performance enhancement and are relatively easy to control. Performance enhancement as indicator of stretch shortening cycle function, is usually measured as an augmentation in concentric performance and force-time kinematics.

The upper body equivalent of the vertical jump testing procedure is that of the pure concentric bench throws (PCBT), and rebound bench throws (RBT) as mentioned in Chapter 1. In the upper body version, Newton et al. (119) found similar force-velocity-power kinetics and kinematics to the vertical jump performance in the lower body, isolated muscles and single-joint movements. This indicates that the control of muscle and performance of upper body explosive movements is very similar to that of the lower body.

The purpose of the two studies in this chapter was therefore to test the repeatability of the testing procedures for determining stretch shortening cycle performance of the lower body (vertical jumps) and upper body (bench throws) using the newly designed and developed NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town).

## **A - THE LOWER BODY**

### **METHODS**

#### **SUBJECTS**

10 Male subjects, between the ages of 20 and 33 years, with various training backgrounds were recruited for this study. Subjects were required to have resistance training experience of at least one year. Their general characteristics are summarized in Table 1. Body fat was measured and expressed as a % (43) and as a sum of 7 skinfolds (129).

**Table 1:** Personal characteristics of subject group (n = 10)

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)	MVC (kg)(*)
26.3 ± 4.4	182.8 ± 6.0	82.8 ± 7.8	60.5 ± 14.4	13.7 ± 3.3	70.6 ± 6.5	252.7 ± 52.4

(\*) Maximum Voluntary Contraction

All subjects were informed about the study and the potential risks involved. The Ethics Committee of the University of Cape Town approved the study. Each subject completed a personal questionnaire regarding medical, training and general information and further signed a written informed consent form complying with the guidelines set by the American College of Sports Medicine, before participating in the study (3).

## **EXPERIMENTAL DESIGN**

The first day was a familiarization day, in which the complete testing procedure was performed as a trial run. This was performed to ensure that the subject knew what was expected of him, and to enhance the reliability of the testing procedure. After the familiarization test, the subjects had to visit the laboratory on a further three occasions where the tests were repeated.

To ensure that the subjects had fully recovered from the previous day's testing, a 48-hour break was given between testing days. Subjects were asked to refrain from any lower body training at least 24 hours before testing and to avoid any heavy or intense training throughout the duration of the study.

On the first day body composition and maximal voluntary contraction (MVC) or isometric leg press were measured. After these tests, the subjects were given the familiarization session, where the various stretch shortening cycle testing procedures were demonstrated. Each subject performed between 3 and 6 trials per test to familiarize himself with the techniques.

The second testing day started with testing for maximal voluntary contraction (MVC) using the leg press unit. This was followed by the stretch shortening cycle tests for the lower body. On the third and fourth testing days, the same testing procedure was repeated. Each day's testing was preceded by a standardized warm-up.

### **TESTING FOR THE MVC ISOMETRIC LEG PRESS**

Before the start of the testing, a standardized generalized warm-up was used consisting of 5 minutes of continuous, self-paced, submaximal shuttle-runs, specific stretches for the lower body and 2 sets of 10 repetition full squats with 20 kg (zero load on the bar). The NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town, Cape Town,



South Africa) was set up and the subject positioned as previously described for the lower body testing in Chapter 2 (p 124). This position was used to create a functional measure of MVC using muscle groups in a similar position to that used during vertical jumping. Each subject was given 3 submaximal warm-up trials, with the instruction to push at an estimated 50, 70 and 90% of maximum respectively against the immovable resistance of the stiffness system. Three maximal trials per subject followed the warm-up trials, with 2 - 3 minutes rest between trials.

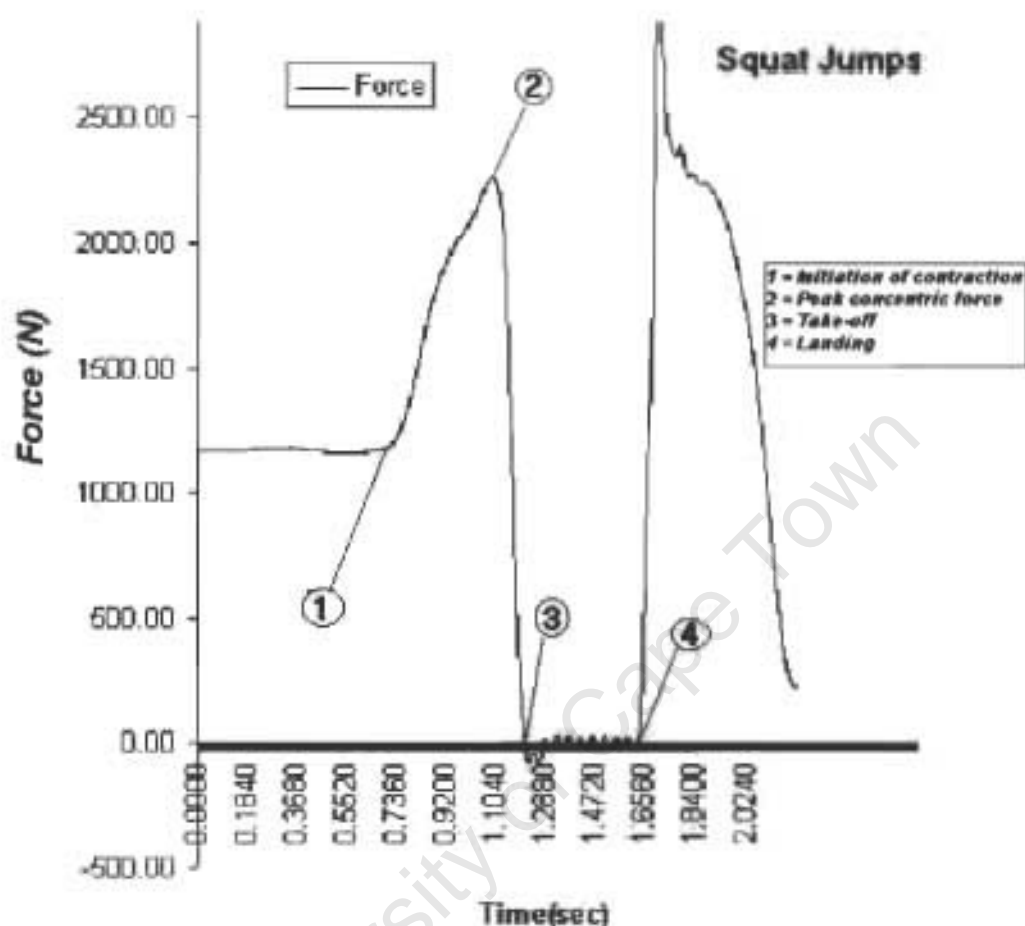
The subjects were instructed to push as fast and as hard as possible, maintaining the contraction for ~5 seconds against the isometric resistance. To ensure that the subjects gave maximal effort in the MVC, the researchers gave vocal encouragement. The maximum isometric force and corresponding integrated EMG obtained from the three trials was recorded and used to normalize the EMG values of the stretch shortening cycle tests that followed. All force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

#### **THE SSC TESTS FOR THE LOWER BODY**

For all the stretch shortening cycle tests, the output from the force plate was set to zero and the sampling strategy for force data remained the same as mentioned above. For the stretch shortening cycle tests of the lower body, the isolated concentric squat jump test and countermovement jump tests were performed. These techniques have previously been described in Chapter 2 (p 124 - 126). Squat jumps were performed first, followed by the countermovement jumps. To minimize the chances of an order effect occurring, the subjects were given as much rest as was necessary to fully recover between trials. The best of three trials for each respective jump were used for further analysis.

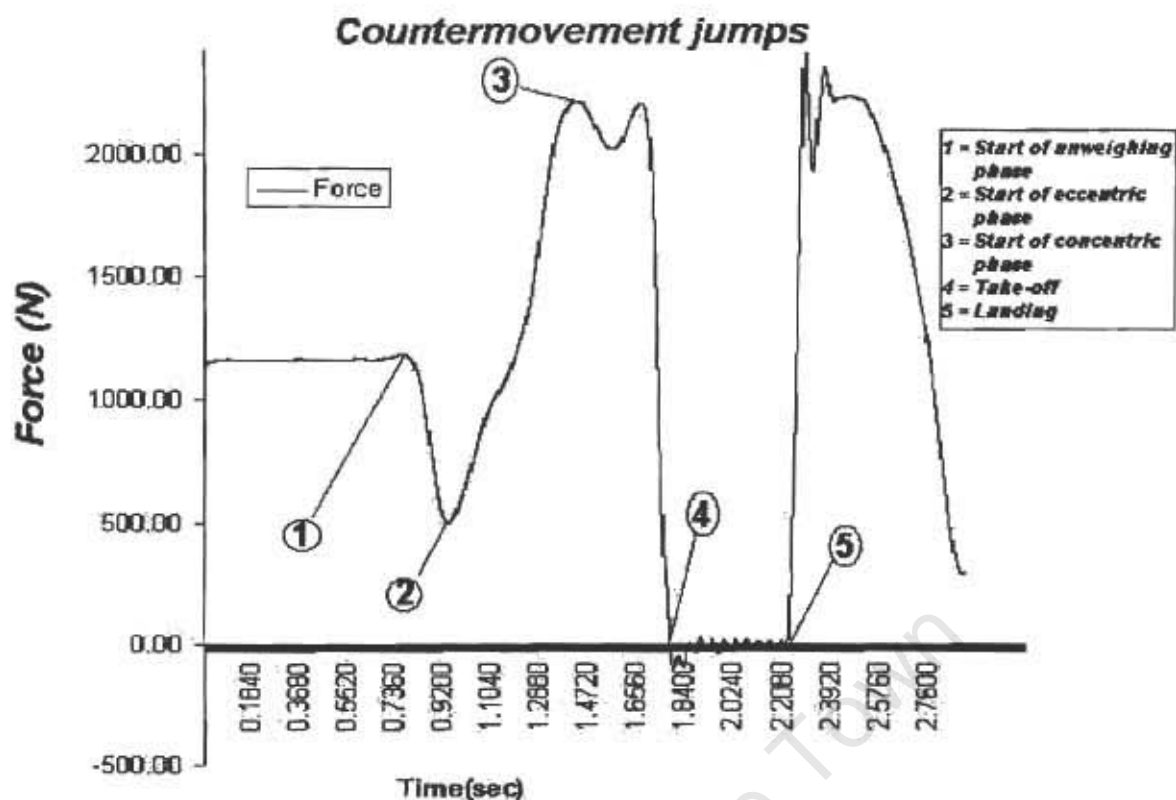
For the analysis of force and EMG, the data were segmented into different regions (Figures 1 and 2). For "Total" measures of squat jumps (Figure 1), the data between initiation of contraction (1) and take-off (3) were used. For the concentric phase of squat jumps, the data between the peak concentric force (2) and take-off (3) were used. It was assumed that the segment between initiation of contraction (1) and peak concentric force (2) was the stretching or tensioning of the series elastic elements of the muscle-tendon

complex before force transmission to the actual jump. Flight time, which is needed for calculation of jump height, was calculated as the time difference between take-off (3) and landing (4).



**Figure 1:** Force-time data of a squat jump. (\* Graph from original data)

In countermovement jumps (Figure 2), there is an un-weighting phase before eccentric contraction where the subject dips downward from a standing position. This phase is essentially a free-falling phase and extends from the start of the un-weighting phase (1) to the start of the eccentric phase (2) where resistance to this down-falling motion is met. For "Total" stretch shortening cycle measures of countermovement jumps, the data between initiation of eccentric contraction (2) and take-off (4) were used. For the concentric phase of countermovement jumps, the data between the peak eccentric force, transition or start of concentric contraction (3) and take-off (4) were used. Flight time, which is needed for calculation of jump height, was calculated as the time difference between take-off (4) and landing (5). All force and EMG data were analysed using this regional breakdown.



**Figure 2:** Force-time data of a countermovement jump. (\* Graph from original data)

Jump height for both tests (squat jumps and countermovement jumps) was calculated using the following formulae:

- Vertical take-off velocities ( $V_v$ ) =  $\frac{1}{2} \times t_{\text{air}}$  (flight time)  $\times g$  (gravitational acceleration or  $9.81 \text{ m.s}^{-2}$ ) (7;25;27;29;101)
- Jump height ( $h$ ) =  $V_v^2/2g$  (101)
- Alternatively one could use ( $h$ ) =  $1.226 \times (t_{\text{air}})^2$  (7)

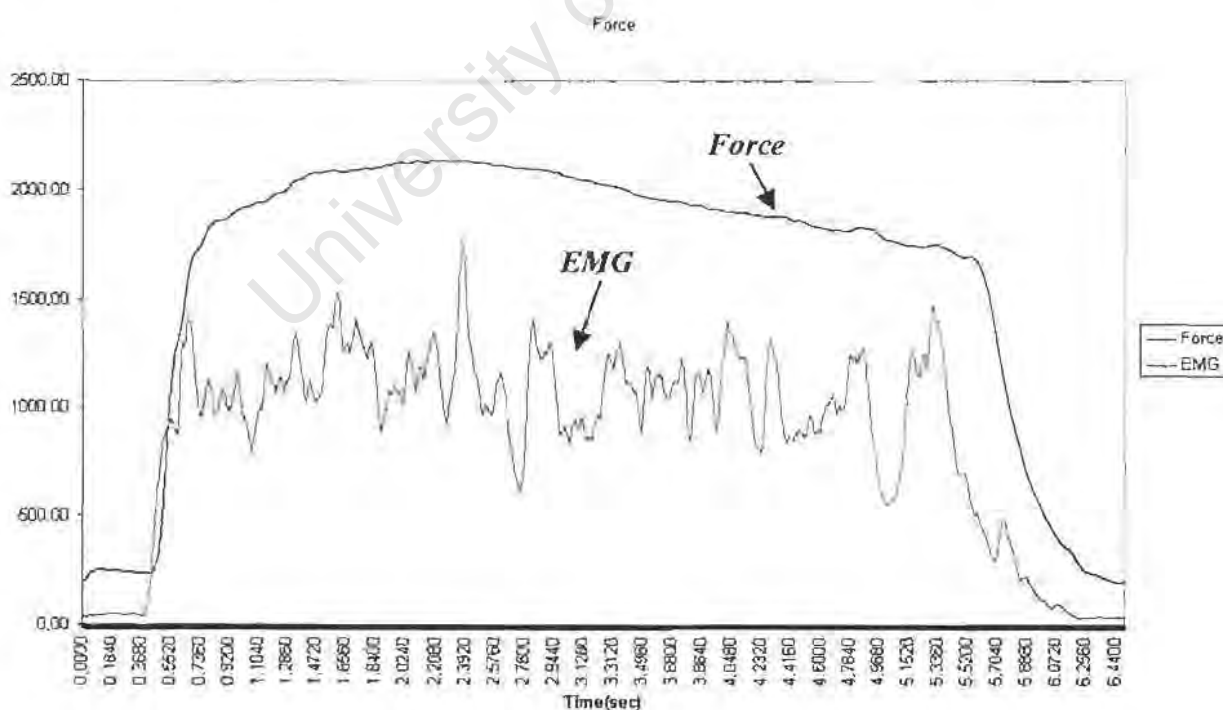
#### ELECTROMYOGRAPHY (EMG)

The vastus medialis oblique muscle (VMO) was used as a measure of neural activation during jumping. Two surface EMG electrodes (Blue sensor SP-00-S, Medicotest A/S, Rugmarken, Denmark) were attached as a pair over the muscle belly with an inter-electrode distance of  $\sim 2 \text{ cm}$  (135) and were placed so that they were orientated parallel with the active muscle fibers. A third neutral reference electrode was placed on the anterior tibia. Electromechanical delay was treated as a systematic error, as it was assumed that this was either negligible or constant (60).

The site for electrode placement was determined as the center of the muscle belly when contracted isometrically, and the electrodes were attached running parallel to the muscle fibers (86). Before the electrodes were attached, the skin surface was prepared. Hair was shaved off using a razor and the skin was scraped with sandpaper to remove the outer layer of epidermal skin cells. The skin surface was then swabbed clean, removing any oils or dirt, using an alcohol swab. Once the alcohol had evaporated, the electrodes were placed on the skin (86).

All raw EMG data were sampled at 2000 Hz using LabVIEW 6.0.2 (National Instruments, Austin, Texas, USA) software and a telemetric EMG system (Noraxon USA, Inc., Scottsdale, Arizona, U.S.A.). Movement artefact was removed using a second-order highpass Butterworth filter with a cut-off frequency of 15 Hz (86). EMG data were then full-wave rectified, and smoothed using a second-order lowpass Butterworth filter with a cut-off frequency of 5 Hz (86), and integrated for further analysis.

3 Seconds of EMG data were extracted from the MVC test data, 1 second after initiation of contraction, and used as the reference MVC IEMG for normalization (Figure 3).



**Figure 3:** An isometric leg press MVC on the NAMS unit showing filtered force, and correspondingly filtered and full-wave rectified EMG data. (\* Graph from original data)

All other EMG data collected were expressed as a percentage of MVC IEMG. To determine if the subjects had exerted a maximum effort during the MVC test on respective days, and to justify comparison of the IEMG (integrated EMG) values, the corresponding 3 seconds of force data were also integrated (MVC impulse) and compared over the three days.

## STATISTICAL ANALYSIS

The data were analysed using the same statistical procedures outlined in Chapter 3 (p 135 - 137).

## RESULTS

Table 2 shows the average data of the various measures used for normalization and in the calculation of stretch shortening cycle performance for each individual measured over the three trials.

**Table 2:** Average test data for the measures in vertical jump testing used in the calculation of stretch shortening cycle performance in the lower extremities for each individual over three trials ( $\bar{X} \pm SD$ ) ( $n = 10$ ).

SUBJECT	MVC peak force (N)	MVC (kg)	MVC impulse (Ns)	MVC IEMG (mVs)	SJ jump height (cm)	CMJ jump height (cm)
SSC <sub>low1</sub>	3068 ± 53	312.7 ± 5.3	8643 ± 144	2883 ± 437	17.1 ± 0.2	18.6 ± 1.5
SSC <sub>low2</sub>	3420 ± 82	348.7 ± 8.4	10011 ± 258	6660 ± 1006	21.2 ± 0.2	24.8 ± 0.2
SSC <sub>low3</sub>	2187 ± 84	223.0 ± 8.6	6335 ± 289	3149 ± 425	21.0 ± 0.6	23.8 ± 2.6
SSC <sub>low4</sub>	2091 ± 164	213.2 ± 16.7	6029 ± 515	2277 ± 1191	14.5 ± 1.5	15.9 ± 0.7
SSC <sub>low5</sub>	2243 ± 227	228.6 ± 23.1	5923 ± 677	2981 ± 355	16.1 ± 0.6	18.1 ± 1.4
SSC <sub>low6</sub>	1805 ± 77	183.9 ± 7.8	5127 ± 268	1882 ± 312	12.5 ± 0.6	13.9 ± 2.0
SSC <sub>low7</sub>	2540 ± 75	258.9 ± 7.7	7397 ± 272	4670 ± 2673	21.8 ± 1.7	24.3 ± 1.1
SSC <sub>low8</sub>	2259 ± 150	230.3 ± 15.3	6479 ± 454	2844 ± 573	17.1 ± 0.2	20.8 ± 0.7
SSC <sub>low9</sub>	2189 ± 11	223.2 ± 1.1	6357 ± 93	8432 ± 912	20.1 ± 0.4	24.5 ± 0.6
SSC <sub>low10</sub>	2991 ± 399	304.9 ± 40.7	8735 ± 1149	2584 ± 111	20.8 ± 0.5	23.5 ± 1.7
Mean	2479 ± 514	252.7 ± 52.4	7104 ± 1548	3836 ± 2127	18.2 ± 3.2	20.8 ± 4.0

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

Table 3 shows the intra-class correlation coefficients and coefficients of variation for the various measures used for normalization and in the calculation of stretch shortening cycle

performance over the three trials. Included are the 95% confidence intervals of both measures of repeatability.

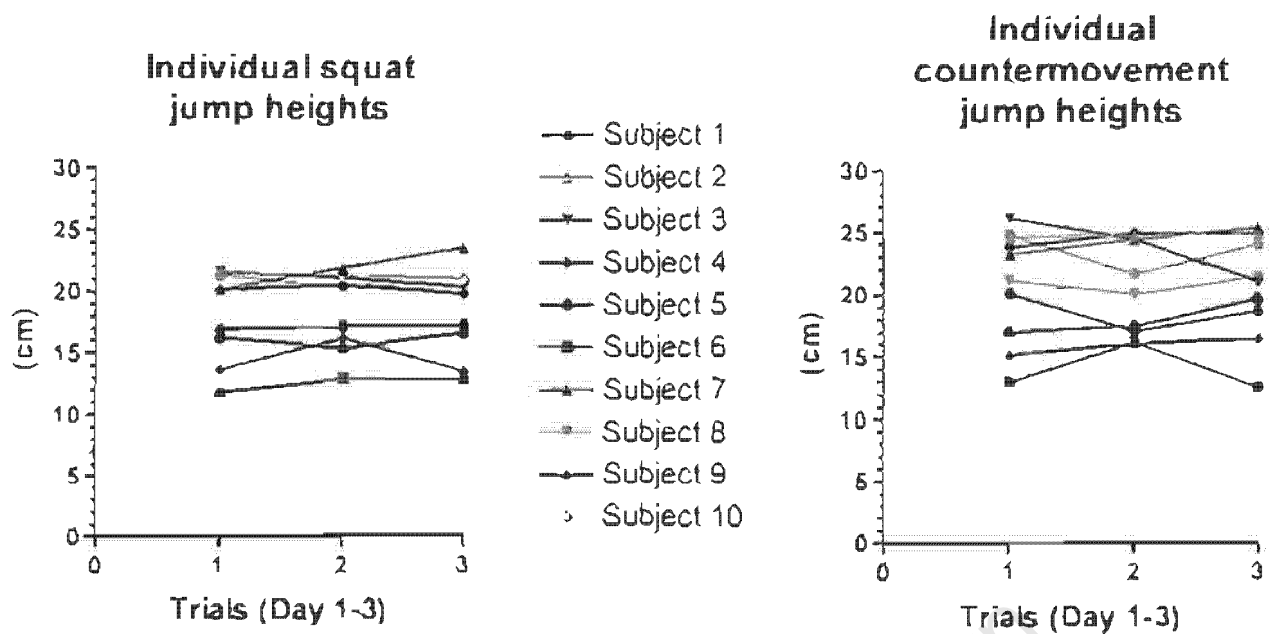
**Table 3:** *Repeatability of measures in vertical jump testing used in the calculation of stretch shortening cycle performance in the lower extremities (n = 10).*

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
MVC peak force (N)	0.96 (0.89-0.99)	5.4 (4.1-8.0)
MVC (kg)	0.96 (0.89-0.99)	5.4 (4.1-8.0)
MVC impulse (Ns)	0.96 (0.89-0.99)	5.9 (4.5-8.7)
MVC IEMG (mVs)	0.92 (0.79-0.98)	21.7 (16.4-32.1)
SJ jump height (cm)	0.98 (0.94-0.99)	3.6 (2.7-5.3)
CMJ jump height (cm)	0.96 (0.89-0.99)	6.4 (4.8-9.5)
SJ MVV (m/s)	0.98 (0.94-0.99)	1.9 (1.4-2.8)
SJ flight time (s)	0.97 (0.92-0.99)	2.2 (1.7-3.3)
CMJ MVV (m/s)	0.96 (0.89-0.99)	3.2 (1.5-3.2)
CMJ flight time (s)	0.95 (0.86-0.99)	3.4 (2.6-5.0)

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, MVV = maximal vertical velocity, IEMG = integrated EMG

Using the definitions of Vincent (146), Table 3 shows that most of these measures are repeatable. For example, all the variables have relatively high intra-class correlation coefficients and low coefficients of variation, except for the MVC IEMG, which has a relatively higher coefficient of variation.

However, because the IEMG-values of the subsequent measures were normalized relative to the maximum IEMG or MVC IEMG, this variability was therefore corrected for. Table 3 and Figure 4 shows that the jump heights, which have been used in the majority of research in this field as indicators of stretch shortening cycle performance, are very repeatable between trials.



**Figure 4:** A graphical representation of the variability for each subject of the concentric jumps (SJ) and the stretch shortening cycle jumps (CMJ) over the three trials.

The stretch shortening cycle jumps or countermovement jumps (CMJ) also appear to be more variable than the concentric jumps or squat jumps (SJ) (Tables 2 and 3, Figure 4).

The one-way ANOVA showed that there were no significant differences between the three trials for the various measures used for normalization and the calculation of stretch shortening cycle performance (Table 4).



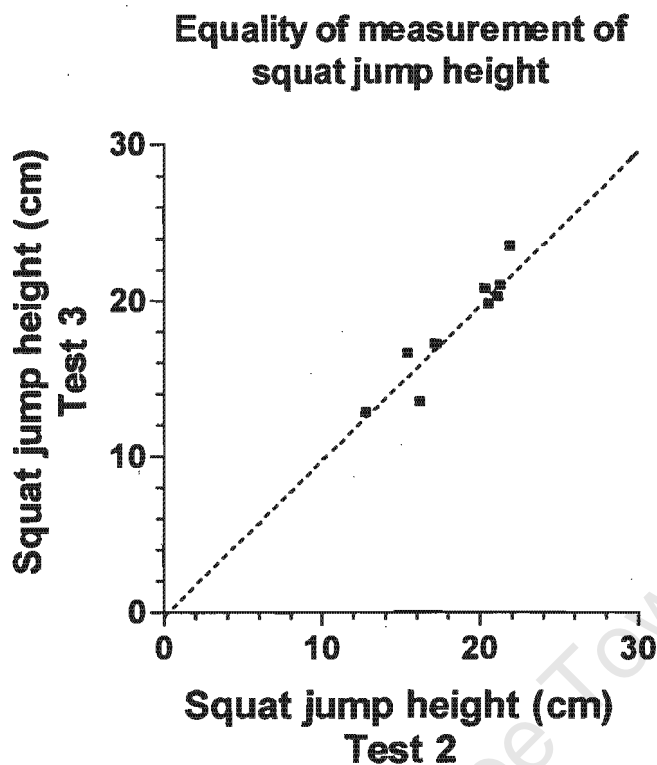
**Table 4:** One-way analysis of variance in measures used in the calculation of stretch shortening cycle performance in the lower extremities (n = 10).

Variable	df effect	MS effect	df error	MS error	F	p-level (*significant)
MVC peak force (N)	2	18084	18	30054	0.60	0.56
MVC (kg)	2	188	18	312	0.60	0.56
MVC impulse (Ns)	2	171063	18	267188	0.64	0.54
MVC IEMG (mVs)	2	2602279	18	971391	2.68	0.10
SJ jump height (cm)	2	0.43	18	0.69	0.62	0.55
CMJ jump height (cm)	2	0.13	18	2.25	0.06	0.94
SJ MVV (m/s)	2	0.001	18	0.002	0.46	0.64
SJ flight time (s)	2	0.00001	18	0.0001	0.09	0.91
CMJ MVV (m/s)	2	0.0002	18	0.006	0.04	0.96
CMJ flight time (s)	2	0.00001	18	0.0003	0.05	0.95

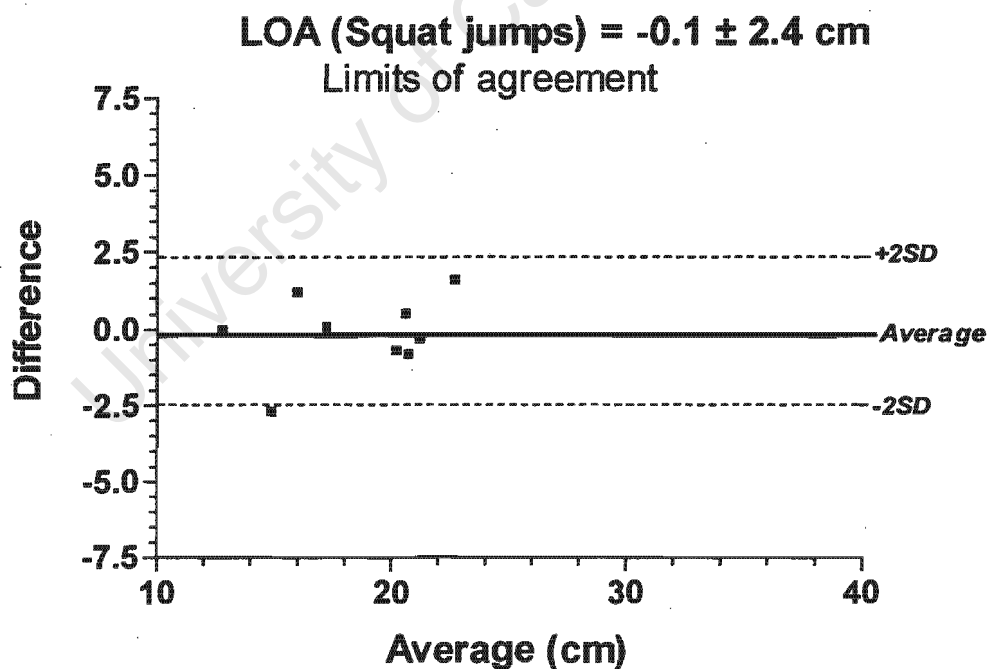
**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG, MVV = Maximal vertical velocity

Figure 5 indicates that the precision (two standard deviations above and below the average difference between trials) by limits of agreement according to Bland-Altman (16) for measurement of squat jump height using the NAMS unit falls within a range of 4.8 cm ( $-0.1 \pm 2.4$  cm).

**A**

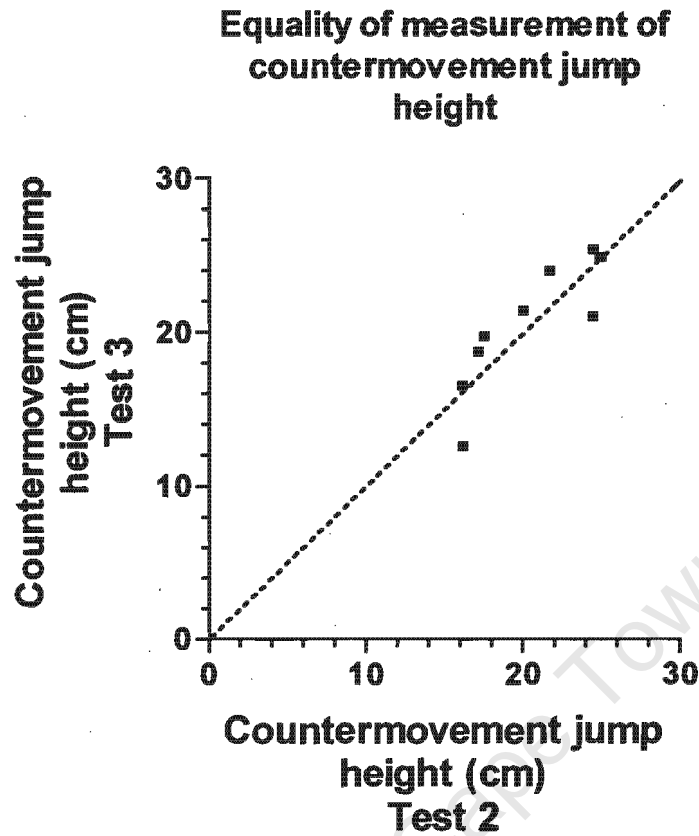


**B**

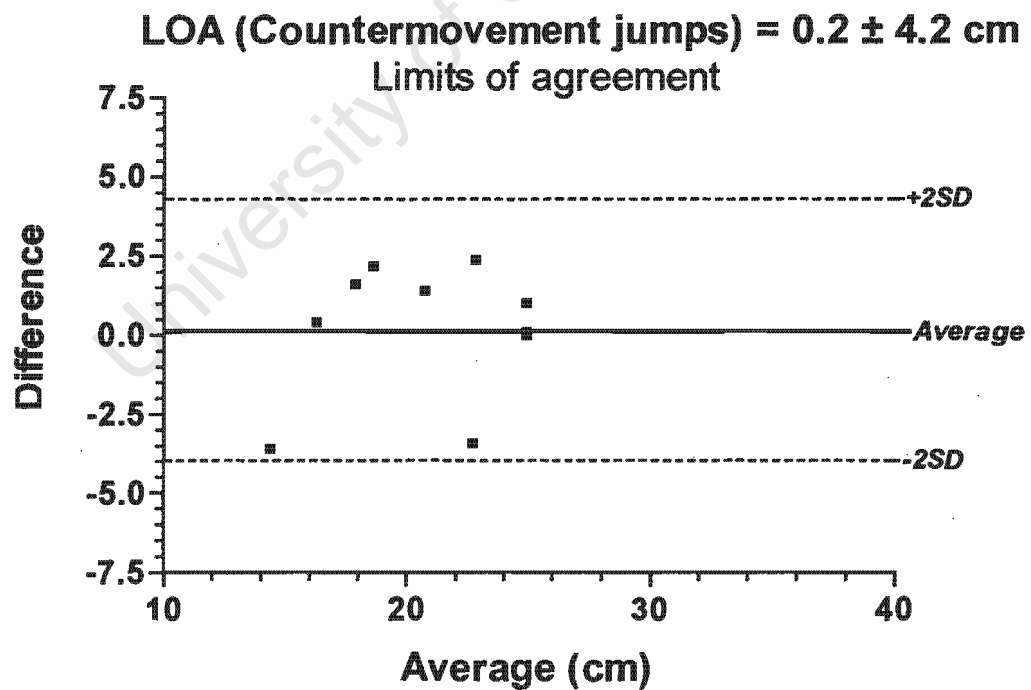


**Figure 5:** A - The equality of measurement for squat jump height between the second and third testing day using the NAMS unit. B – The limits of agreement for measurement of squat jump height between the second and third testing day using the method of Bland-Altman (16).

**A**



**B**



**Figure 6:** A - The equality of measurement for countermovement jump height between the second and third testing day using the NAMS unit. B – The limits of agreement for measurement of countermovement jump height between the second and third testing day using the method of Bland-Altman (16).

Figure 6 indicates that the precision of measurement of countermovement jump height using the NAMS unit, defined by the limits of agreement (two standard deviations above and below the average difference between trials) falls within a range of 8.4 cm ( $0.2 \pm 4.2$  cm) (16).

Table 5 shows the average data for each subject for the "Total" force and -EMG measures used in stretch shortening cycle testing of the lower extremities over the three trials.

**Table 5:** Average test data for "Total" force and -EMG measures used in stretch shortening cycle testing of the lower extremities ( $n = 10$ ).

SUBJECT	"Total" SJ impulse (Ns)	"Total" SJ IEMG (%MVC)	"Total" CMJ impulse (Ns)	"Total" CMJ IEMG (%MVC)
SSClow1	645.7 $\pm$ 21.3	16.0 $\pm$ 3.7	992.1 $\pm$ 34.4	25.2 $\pm$ 0.7
SSClow2	824.6 $\pm$ 27.2	11.8 $\pm$ 1.0	1225.3 $\pm$ 69.5	16.8 $\pm$ 1.8
SSClow3	659.7 $\pm$ 6.3	12.2 $\pm$ 1.7	1030.0 $\pm$ 5.2	22.1 $\pm$ 0.3
SSClow4	655.8 $\pm$ 27.3	16.7 $\pm$ 2.2	890.5 $\pm$ 21.9	19.9 $\pm$ 0.4
SSClow5	700.9 $\pm$ 12.5	12.2 $\pm$ 1.0	1106.3 $\pm$ 25.7	22.0 $\pm$ 1.9
SSClow6	652.4 $\pm$ 23.2	31.7 $\pm$ 3.8	942.9 $\pm$ 60.4	43.0 $\pm$ 5.8
SSClow7	748.3 $\pm$ 3.0	11.7 $\pm$ 3.6	1091.3 $\pm$ 25.0	15.0 $\pm$ 4.0
SSClow8	597.8 $\pm$ 42.7	17.2 $\pm$ 2.7	863.7 $\pm$ 57.6	26.7 $\pm$ 4.4
SSClow9	595.4 $\pm$ 43.3	5.0 $\pm$ 0.3	950.8 $\pm$ 6.5	12.8 $\pm$ 1.5
SSClow10	653.6 $\pm$ 45.4	9.4 $\pm$ 1.6	987.0 $\pm$ 23.0	15.2 $\pm$ 2.1
Mean	673.4 $\pm$ 69.2	14.4 $\pm$ 2.1	1008.0 $\pm$ 109.1	21.9 $\pm$ 8.7

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

During testing on one of the trials, the telemetric EMG did not function properly during the jump tests and all the subject's EMG data on this specific day were discarded and not used in the ensuing calculations. These measurements are indicated by <sup>#</sup>( $n = 9$ ) in Tables 6, 7, 10, 11 and 12.

Table 6 shows the intra-class correlation coefficients and coefficients of variation for the "Total" force and -EMG measures used in stretch shortening cycle testing of the lower extremities over the three trials. Included are the 95% confidence intervals of the both measures of repeatability.

**Table 6:** Repeatability of "Total" force and -EMG measures used in stretch shortening cycle testing of the lower extremities (n = 10).

Variable #(n = 9)	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
"Total" SJ impulse (Ns)	0.94 (0.84-0.98)	3.9 (3.0-5.8)
"Total" SJ IEMG (%MVC) #	0.96 (0.88-0.99)	14.7 (11.0-22.4)
"Total" CMJ impulse (Ns)	0.96 (0.89-0.99)	3.3 (2.5-4.9)
"Total" CMJ IEMG (%MVC) #	0.96 (0.88-0.99)	10.8 (8.0-16.4)

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

Table 6 shows that these measures are repeatable with relatively high intra-class correlation coefficients and low coefficients of variation. Once again the EMG related variables have the highest variation. However, the variation of these EMG-related measures is still within acceptable limits (146).

**Table 7:** One-way analysis of variance in "Total" force and -EMG measures used in stretch shortening cycle testing of the lower extremities (n = 10).

Variable #(n = 9)	df effect	MS effect	df error	MS error	F	p-level (*significant)
"Total" SJ impulse (Ns)	2	1788.32	18	739.85	2.42	0.12
"Total" SJ IEMG (%MVC) #	2	21.37	16	4.22	5.06	0.02*
"Total" CMJ impulse (Ns)	2	39.86	18	1701.35	0.02	0.98
"Total" CMJ IEMG (%MVC) #	2	14.33	16	8.68	1.65	0.22

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

The one-way ANOVA showed significant differences over the three trials for the "Total" SJ IEMG of the squat jump test (Table 7). A Scheffe's post-hoc test performed on these data showed a significant difference between trials 1 and 3 ( $p < 0.03$ ) (Table 8).

**Table 8:** Scheffe's Post-hoc test to analyse differences between trials in the lower body stretch shortening cycle tests ( $n = 10$ ).

Variable: "Total" SJ IEMG (*significant)			
Testing day	Average day 1: 15.87 %MVC	Average day 2: 13.63 %MVC	Average day 3: 12.91 %MVC
1		0.10	0.03*
2	0.10		0.8
3	0.03*	0.8	

**Index:** SJ = squat jumps, IEMG = integrated EMG

Table 9 shows the average data for each subject for the concentric force and -EMG measures used in stretch shortening cycle testing of the lower extremities over the three trials.

**Table 9:** Average test data for concentric force and -EMG measures in stretch shortening cycle testing of the lower extremities ( $n = 10$ ).

SUBJECT	SJ concentric impulse (Ns)	SJ concentric IEMG (%MVC)	CMJ concentric impulse (Ns)	CMJ concentric IEMG (%MVC)
SSClow1	199.2 $\pm$ 11.7	5.2 $\pm$ 1.1	644.4 $\pm$ 58.7	16.4 $\pm$ 1.6
SSClow2	169.4 $\pm$ 9.6	2.5 $\pm$ 0.2	691.1 $\pm$ 39.8	9.3 $\pm$ 0.7
SSClow3	144.4 $\pm$ 6.0	3.2 $\pm$ 0.5	545.9 $\pm$ 12.5	10.3 $\pm$ 1.8
SSClow4	158.8 $\pm$ 21.2	4.8 $\pm$ 0.5	443.2 $\pm$ 49.5	7.0 $\pm$ 6.1
SSClow5	155.8 $\pm$ 5.1	2.9 $\pm$ 0.4	472.2 $\pm$ 39.5	10.0 $\pm$ 0.9
SSClow6	191.1 $\pm$ 14.6	11.6 $\pm$ 1.2	454.2 $\pm$ 126.1	25.7 $\pm$ 3.4
SSClow7	168.2 $\pm$ 10.9	2.5 $\pm$ 0.9	712.5 $\pm$ 86.2	10.0 $\pm$ 3.1
SSClow8	157.6 $\pm$ 4.1	6.6 $\pm$ 1.6	484.1 $\pm$ 53.0	14.5 $\pm$ 4.4
SSClow9	195.7 $\pm$ 42.7	1.0 $\pm$ 0.5	531.4 $\pm$ 52.7	3.9 $\pm$ 1.6
SSClow10	212.1 $\pm$ 23.4	3.2 $\pm$ 0.4	565.1 $\pm$ 6.7	8.9 $\pm$ 1.4
Mean	175.2 $\pm$ 22.6	4.3 $\pm$ 3.0	554.4 $\pm$ 98.1	11.6 $\pm$ 6.1

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

Table 10 shows the intra-class correlation coefficients and coefficients of variation for the concentric force and -EMG measures used in stretch shortening cycle testing of the lower extremities over the three trials. Included are the 95% confidence intervals of both the measures of repeatability.

**Table 10:** Repeatability of concentric force and -EMG measures in stretch shortening cycle testing of the lower extremities (n = 10).

Variable *(n=9)	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
SJ concentric impulse (Ns)	0.77 (0.48-0.93)	8.2 (6.2-12.1)
SJ concentric IEMG (%MVC) *	0.98 (0.94-1.00)	20.7 (15.4-31.5)
CMJ concentric impulse (Ns)	0.87 (0.67-0.96)	9.9 (7.5-14.6)
CMJ concentric IEMG (%MVC) *	0.95 (0.85-0.99)	26.1 (19.4-39.7)

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

It is noticeable that the measurements in Tables 9 and 10 have a higher variability and lesser repeatability than the measurements in the previous data tables (Tables 2, 3, 5 and 6). This is evident in the greater confidence interval ranges of intra-class correlation coefficients and coefficients of variation. The one-way ANOVA showed that there were no significant differences in the concentric force and -EMG measures used in stretch shortening cycle testing of the lower extremities over the three trials (Table 11).

**Table 11:** One-way analysis of variance in concentric force and -EMG measures in stretch shortening cycle testing of the lower extremities (n = 10).

Variable *(n = 9)	df effect	MS effect	df error	MS error	F	p-level (*significant)
SJ concentric impulse (Ns)	2	36.90	18	380.07	0.10	0.91
SJ concentric IEMG (%MVC) *	2	1.83	16	0.63	2.92	0.08
CMJ concentric impulse (Ns)	2	2025.37	18	4015.04	0.50	0.61
CMJ concentric IEMG (%MVC) *	2	1.64	16	6.24	0.26	0.77

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG

Table 12 shows the intra-class correlation coefficients, coefficients of variation and respective 95% confidence intervals of all the possible mathematical derivatives that could be used to calculate or measure stretch shortening cycle potentiation of performance in



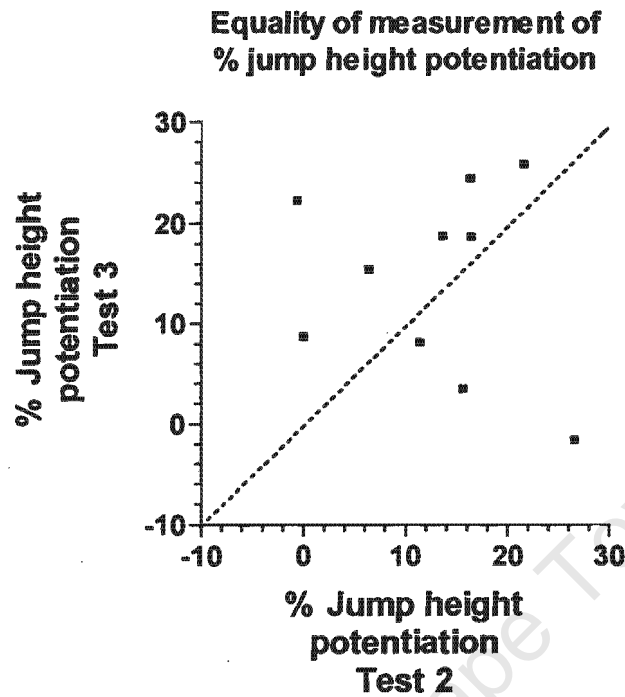
the lower body musculature. Table 12 also includes the results of the one-way ANOVA for significant difference between trials, with significance accepted as  $p < 0.05$ .

**Table 12:** Repeatability of mathematical derivatives of vertical jump performance measures in stretch shortening cycle testing of the lower extremities ( $n = 10$ ).

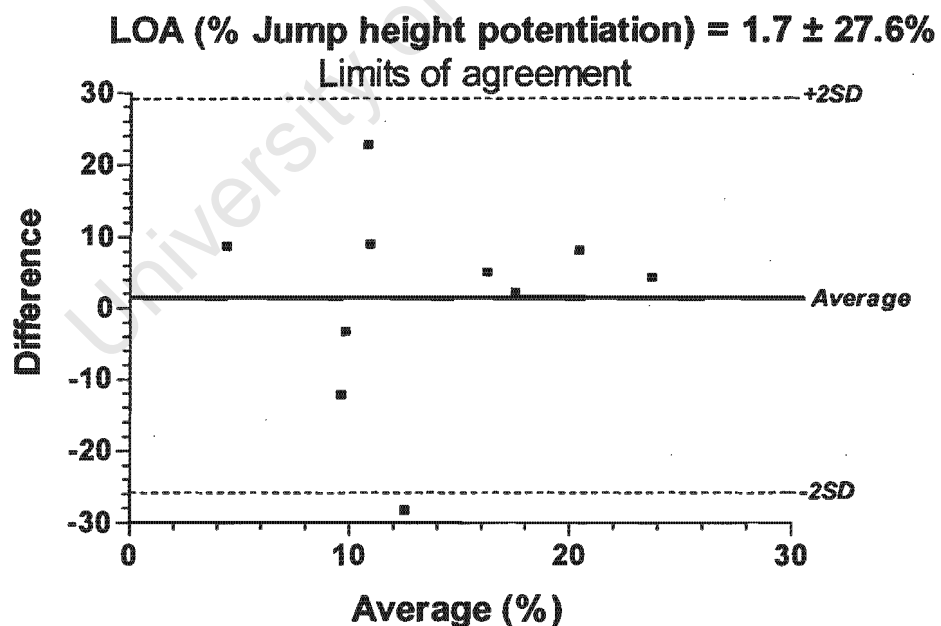
Variable * ( $n = 9$ )	Intra-class Correlation Coefficient (95% Confidence Interval)	% Coefficient of Variation (95% Confidence Interval)	p-level (* significant difference between trials)
"Total" SJ F:EMG ratio (N/%MVC) *	0.98 (0.94-1.00)	12.6 (9.4-19.2)	0.05*
"Total" CMJ F:EMG ratio (N/%MVC) *	0.96 (0.88-0.99)	9.5 (7.1-14.5)	0.44
Concentric SJ F:EMG ratio (N/%MVC) *	0.94 (0.83-0.98)	19.1 (14.2-29.1)	0.97
Concentric CMJ F:EMG ratio (N/%MVC) *	0.95 (0.85-0.99)	14.0 (10.4-21.3)	0.89
"Total" CMJ/"Total" SJ impulse potentiation ratio	0.52 (0.13-0.83)	5.2 (3.9-7.7)	0.34
CMJ/SJ concentric impulse potentiation ratio	0.89 (0.72-0.97)	10.8 (8.2-16.0)	0.67
CMJ-SJ Height difference (cm)	0.48 (0.09-0.81)	57.8 (43.7-85.5)	0.59
CMJ-SJ height % Potentiation	-0.08 (-0.32-0.39)	57.8 (43.7-85.5)	0.73
CMJ/SJ height potentiation ratio	-0.08 (-0.32-0.39)	6.3 (4.8-9.3)	0.73
"Total" CMJ-SJ F:EMG difference (N/%MVC) *	0.94 (0.83-0.98)	124.1 (92.4-188.9)	0.47
"Total" CMJ-SJ F:EMG % difference *	0.82 (0.55-0.95)	146.7 (109.3-223.3)	0.39
"Total" CMJ F:EMG/"Total" SJ F:EMG ratio *	0.82 (0.55-0.92)	10.2 (7.6-15.5)	0.39
CMJ-SJ concentric F:EMG difference (N/%MVC) *	0.93 (0.80-0.98)	442.6 (329.6-673.6)	0.98
CMJ-SJ concentric F:EMG % difference *	0.82 (0.55-0.95)	67.1 (50.0-102.1)	0.31
Concentric CMJ F:EMG/ Concentric SJ F:EMG ratio *	0.82 (0.55-0.95)	14 (10.4-21.3)	0.31

**Index:** MVC = maximal voluntary contraction, SJ = squat jumps, CMJ = countermovement jumps, EMG = integrated EMG, F = integrated force

**A**



**B**



**Figure 7:** A - The equality of measurement for jump height potentiation via the stretch shortening cycle between the second and third testing day using the NAMS unit. B – The limits of agreement for measurement of jump height potentiation between the second and third testing day using the method of Bland-Altman (16).

Figure 7 indicates that the precision (two standard deviations above and below the average difference between trials) by limits of agreement according to Bland-Altman (16) for measurement of jump height potentiation using the NAMS unit falls within a range of 55.2 % ( $1.7 \pm 27.6$  %).

From the Table 12, it can clearly be seen that the most repeatable derivative measures are those of the "Total" SJ F: EMG-, "Total" CMJ F: EMG-, Concentric SJ F: EMG- and Concentric CMJ F: EMG ratios and their derivatives ("Total" CMJ F: EMG/"Total" SJ F: EMG- and Concentric CMJ F: EMG/Concentric SJ F: EMG ratios). A similar pattern was visible in the original data where the "Concentric-defined" measures were less repeatable than the "Total"-defined measures. The remainder of the mathematical derivatives all lack significant repeatability to be of practical relevance in future studies. Of great importance is the fact that the frequently used measure of stretch shortening cycle potentiation i.e. CMJ-SJ % height potentiation was not repeatable. This is visible in Figure 7, where the precision of measure (16) is also indicated. Following the one-way ANOVA, the "Total" SJ F: EMG ratio showed a significant difference between the three days ( $p < 0.05$ ) and a Scheffé's Post-hoc test was performed to identify where the difference lay (Table 13).

**Table 13:** Scheffe's Post-hoc test to analyse differences between trials in the lower body stretch shortening cycle tests ( $n = 10$ ).

Variable: "Total" SJ F: EMG ratio (*significant)			
Testing day	Average day 1: 55.86 N/%MVC	Average day 2: 60.46 N/%MVC	Average day 3: 63.63 N/%MVC
1		0.30	0.05*
2	0.30		0.56
3	0.05*	0.56	

**Index:** SJ = squat jumps, EMG = integrated EMG, F = integrated force

The Scheffe's post-hoc test performed on these data showed a significant difference between trials 1 and 3 ( $p < 0.05$ ) (Table 13).

## DISCUSSION

The main aim of this study was to test, using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town), whether the various measures that are used in

testing for the stretch shortening cycle performance of the lower body musculature were repeatable.

The first main finding of this study is that the jump height measures of using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town) were repeatable over three different testing days (Table 3). These are the main measures to be used in future research using the NAMS unit in differentiating stretch shortening cycle performance amongst different individuals.

The static or squat jumps showed an intra-class correlation coefficient of  $R = 0.98$  (95% CI: 0.94-0.99) and coefficient of variation of  $CV = 3.6$  (95% CI: 2.7-5.3)%. The stretch shortening cycle or countermovement jumps showed an intra-class correlation coefficient of  $R = 0.96$  (95% CI: 0.89-0.99) and coefficient of variation of 6.4 (95% CI: 4.8-9.5)%. These findings compare favourably with previous research. Goodwin et al. (63) showed an intra-class correlation coefficient of  $R = 0.96$  for vertical jump height. Arteaga et al. (6) showed pooled coefficients of variation of 5.4% and 6.3% over six days' testing for squat jumps and countermovement jumps respectively. Moir et al. (114) tested the reliability of squat jumps and countermovement in an unloaded, and loaded (with a 10 kg weighted jacket), over a period of five days. They reported intra-class correlations coefficients of  $r = 0.89 - 0.95$  and coefficients of variation of  $CV = 2.1 - 2.6$  %. Although both measures (Table 3) were repeatable, the countermovement jumps seemed to be more variable than the squat jumps. Some subjects needed more time to become familiarized with the countermovement jumps than other subjects. However with the squat jumps this problem did not occur.

The main problem the subjects had learning the countermovement jumps was with the standardization of the depth of the jumps. As mentioned in the literature review in Chapter 1, the selection of jumping techniques varies between subjects (21). This was confirmed during the familiarization trials, where some subjects instinctively selected smaller ranges of motion for the depth of their countermovement jumps, whereas other subjects selected larger ranges of motion. Subjects had to be trained to override their natural instinct and to react to the sound trigger, which controlled and standardized the depth of the countermovement jumps. This raises the question whether standardizing the depth of the jump for research purposes is appropriate, as the standardized depth may not

be the most efficient depth for every subject. This might also explain why certain subjects had more difficulty familiarizing themselves with the countermovement jumps and also explains the increased variability in the measurement when compared to the squat jumps. However, both measures still had acceptable repeatability for the measurement to be considered reliable (146), and can be used with confidence in future research using the NAMS unit. A recommendation however is that certain individuals may need more coaching when learning the countermovement jumps using the NAMS unit to improve the accuracy of the measure. This needs to be considered in future studies when the testing protocol is planned.

The precision of measurement of the jump height data is another factor that needs to be considered for future testing. Determined by the criteria of Bland-Altman (16), the precision of measurement of the jump height data, for squat jumps falls within a range of 4.8 cm and for countermovement jumps falls within a range of 8.4 cm (Figures 5 and 6).

The second important finding in this study was that the peak force (N) and peak load (kg) measures ( $R = 0.96$  (95% CI: 0.89-0.99);  $CV = 5.4$  (95% CI: 4.1-8.0)% for both measurements) of the MVC leg-press test and the MVC impulse ( $R = 0.96$  (95% CI: 0.89-0.99);  $CV = 5.9$  (95% CI: 4.5-8.7)%) (Table 3) as a control measure of maximal effort were also very repeatable. From these data one can conclude that the subjects gave maximal efforts during the test on each of the 3 days.

The third finding of this study was that MVC IEMG data (Table 3), although reasonably repeatable, was more variable than the other force measures and jump height calculations. This could be due to minor positional differences of the electrodes during the preparation of the subject, subcutaneous fat, or electrical conductivity of the skin surface. It did nonetheless compare favourably with previous research. Isear et al. (82) showed intra-class correlations of  $R = 0.89-0.98$  for the MVC EMG data of the various muscle groups they tested. However, due to the fact that the IEMG data of the jump trials were normalized against the respective maximum recruitment or MVC IEMG of the day, this variation was corrected for.

The fourth major finding of this study comparing the "Total" force and -IEMG measures (Table 6) was that a similar pattern of repeatability was apparent to the vertical jump

performance measures (Table 3). The force-time measures i.e. *"Total"* SJ impulse ( $R = 0.94$  (95% CI: 0.84-0.98);  $CV = 3.9$  (95% CI: 3.0-5.8)) and *"Total"* CMJ impulse ( $R = 0.96$  (95% CI: 0.89-0.99);  $CV = 3.3$  (95% CI: 2.5-4.9)) were repeatable. The EMG measures i.e. *"Total"* SJ IEMG ( $R = 0.96$  (95% CI: 0.88-0.99);  $CV = 14.7$  (95% CI: 11.0-22.4)) and *"Total"* CMJ IEMG ( $R = 0.96$  (95% CI: 0.88-0.99);  $CV = 10.8$  (95% CI: 8.0-16.4)) were also repeatable, but they were however more variable than the force-time measures. This could possibly be due to different recruitment patterns during the jumps, or as mentioned earlier, minor electrode positional differences, and differences in subcutaneous fat-, and skin surface electrical conductivity.

The post-hoc results of Table 8 show that there was a progressive reduction in the group *"Total"* SJ IEMG over the three days. This could mean that the group became more efficient i.e. there was a learning effect in muscle recruitment over the three days. The fact that this effect did not show in the countermovement jumps and the fact that there was no significant change in the heights jumped over the three days, questions this reasoning and suggests that this finding is representative of a type 1 error, i.e. showing differences where there are not any. The post-hoc results of Table 13 mirror the changes in Table 8. They show a progressive increase in *"Total"* SJ F: EMG over the three days. This seems to be an artefact of the significant reduction ( $p < 0.03$ ) in *"Total"* SJ IEMG (Table 8), as *"Total"* SJ IEMG forms the one part of the *"Total"* SJ F: EMG ratio. In Table 7 there is no significant difference in *"Total"* SJ Impulse ( $p < 0.12$ ) over the three days, which confirms this interpretation as *"Total"* SJ Impulse forms the other part of the ratio.

The fifth finding of this study comparing the concentric force and -IEMG measures (Table 10) was that the force-time measures i.e. SJ concentric impulse ( $R = 0.77$  (95% CI: 0.48-0.93);  $CV = 8.2$  (95% CI: 6.2-12.1)) and CMJ concentric impulse ( $R = 0.87$  (95% CI: 0.67-0.96);  $CV = 9.9$  (95% CI: 7.5-14.6)) were less repeatable than their respective *"Total"* measures (Table 6). The EMG measures i.e. SJ concentric IEMG ( $R = 0.98$  (95% CI: 0.94-1.00);  $CV = 20.7$  (95% CI: 15.4-31.5)) and CMJ concentric IEMG ( $R = 0.95$  (95% CI: 0.85-0.99);  $CV = 26.1$  (95% CI: 19.4-39.7)) were also more variable than the force-time measures and their respective *"Total"* IEMG measures (Table 6). This could possibly be due to different technique or recruitment patterns during the jumps, or as mentioned earlier minor electrode positional differences, subcutaneous fat, and skin surface electrical conductivity. Based on these findings, for comparison between

concentric and stretch shortening cycle measures, one would not recommend use of the concentric-defined measures (Table 10) unless a larger number of subjects were used.

The sixth important finding of this study was that the mathematical derivatives (Table 12), which are often used to calculate stretch shortening cycle potentiation, are not repeatable. Various researchers throughout the years have defined stretch shortening cycle potentiation of the muscle-tendon complex as the difference between squat jump height and countermovement jump height expressed as a percentage (6;73;97;101;151;152):

$$\text{Percentage Difference (Stretch shortening cycle performance)} = (\text{CMJ-SJ})/\text{SJ} \times 100\%$$

However, based on the data in this study, this index is not a reliable measure of stretch shortening cycle potentiation, as it is not repeatable. Even though the individual measures of squat jumps ( $R = 0.98$  (95% CI: 0.94-0.99);  $CV = 3.6$  (95% CI: 2.7-5.3)) and countermovement jumps ( $R = 0.96$  (95% CI: 0.89-0.99);  $CV = 6.4$  (95% CI: 4.8-9.5)) were repeatable (Table 3), the mathematical derivative i.e. percentage jump height potentiation ( $R = -0.08$  (95% CI: -0.32-0.39);  $CV = 57.8$  (95% CI: 43.7-85.5)) was not (Table 12). Arteaga et al. (6) showed a pooled coefficient of variation of 84% in this measure, using a slightly different methodology. However, as this measure is always a relative measure, it would make sense that if for example someone was tested on two different occasions, and he jumped slightly higher on his squat jump and slightly lower on his countermovement jump on the second occasion than the first testing day, the percentage difference would differ significantly. Our data (Table 12, Figure 7) confirms this. Therefore we have to question the efficacy of this index as a reliable measure of stretch shortening cycle potentiation. Of the other derived measures, the most reliable measure for gauging stretch shortening cycle performance would seem to be the ratio of "Total" CMJ F: EMG/"Total" SJ F: EMG ( $R = 0.82$  (95% CI: 0.55-0.92);  $CV = 10.2$  (95% CI: 7.6-15.5)) even though it is not a functional measure of performance. But one has to again question the use of that ratio as well, because if you express the same relationship between the same data comprising the ratio differently e.g. as an absolute difference (subtracting the one from the other) ( $R = 0.94$  (95% CI: 0.83-0.98);  $CV = 124.1$  (95% CI: 92.4-188.9)) or as a percentage change (the absolute difference expressed as a percentage change) ( $R = 0.82$  (95% CI: 0.55-0.95);  $CV = 146.7$  (95% CI: 109.3-223.3)) their variability increases substantially. The measures of "Total" SJ F: EMG ( $R = 0.98$  (95% CI: 0.94-1.00);  $CV = 12.6$  (95% CI:



9.4-19.2) and "Total" CMJ F: EMG ( $R = 0.96$  (95% CI: 0.88-0.99);  $CV = 9.5$  (95% CI: 7.1-14.5) are however, very repeatable as individual measures.

## CONCLUSION

The first main conclusion of this study is that the measurements of stretch shortening cycle performance of the lower body are generally repeatable when measured with the NAMS unit (Zest Manufacturing, PTY (Ltd) and University of Cape Town). However, the concentric-defined measurements are less repeatable and should be used with caution.

The second main conclusion of this study is that even though many of the actual data or individual measurements of stretch shortening cycle performance are repeatable as individual entities, when they are incorporated into ratios and mathematical derivatives, they are no longer necessarily repeatable or reliable.

## B - THE UPPER BODY

## METHODS

### SUBJECTS

10 Male subjects, between the ages of 21 and 33 years and with various training backgrounds were recruited for this study. Subjects were required to have resistance training experience of at least one year. Their general characteristics are summarized in Table 14. Body fat was measured and expressed as a % (43) and as a sum of 7 skinfolds (129).

**Table 14:** Personal characteristics of subject group ( $n = 10$ )

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)	MVC (kg)(*)
26.9 ± 4.2	181.7 ± 6.4	83.3 ± 9.8	77.1 ± 29.8	15.9 ± 4.5	70.3 ± 9.1	88.6 ± 40.5

(\*) Maximum Voluntary Contraction

All subjects were informed about the study and the potential risks involved. The Ethics Committee of the University of Cape Town approved the study. Each subject completed a personal questionnaire regarding medical, training and general information and further signed a written informed consent form complying with the guidelines set by the American College of Sports Medicine, before participating in the study (3).

## **EXPERIMENTAL DESIGN**

The first day was a familiarization day, in which the complete testing procedure was performed as a trial run. This was performed to ensure that the subject knew what was expected of him, and to enhance the reliability of the testing procedure. After the familiarization test, the subjects had to visit the laboratory on a further three occasions where the tests were repeated.

To ensure that the subjects had fully recovered from the previous day's testing, a 48-hour break was given between testing days. Subjects were asked to avoid any upper body training at least 24 hours before testing and to refrain from any heavy or intense training throughout the duration of the study.

On the first day body composition and maximal voluntary contraction (MVC) or isometric bench press were measured. After these tests, the subjects were given the familiarization session, where the various stretch shortening cycle testing procedures were demonstrated. Subjects performed between 3 and 6 trials per test to familiarize themselves.

The second testing day started with testing for maximal voluntary contraction (MVC) using the NAMS unit. This was followed by the bench throw tests for the upper body. On the third and fourth testing days, the same testing procedure was repeated. Each day's testing was preceded by a standardized warm-up.

### **TESTING FOR THE MVC ISOMETRIC BENCH PRESS**

Before the start of testing a standardized generalized warm-up was used consisting of 5 minutes of continuous, self-paced, submaximal shuttle-runs, 2 sets of 10 push-ups with body mass as resistance, specific stretches for the upper body and 1 set of 10 repetitions

bench press on the NAMS unit with 21.32 kg (zero load on the bar). The NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town, Cape Town, South Africa) was set up and the subject positioned as previously described for the upper body in Chapter 2 (p 124). This position was used to create a functional measure of MVC using muscle groups in a similar position to that used during bench throws. Each subject was given 3 submaximal warm-up trials, where they were instructed to push at an estimated 50, 70 and 90% of maximum respectively against the immovable resistance of the NAMS unit. Three maximal trials per subject followed the warm-up, with 2 - 3 minutes rest between trials.

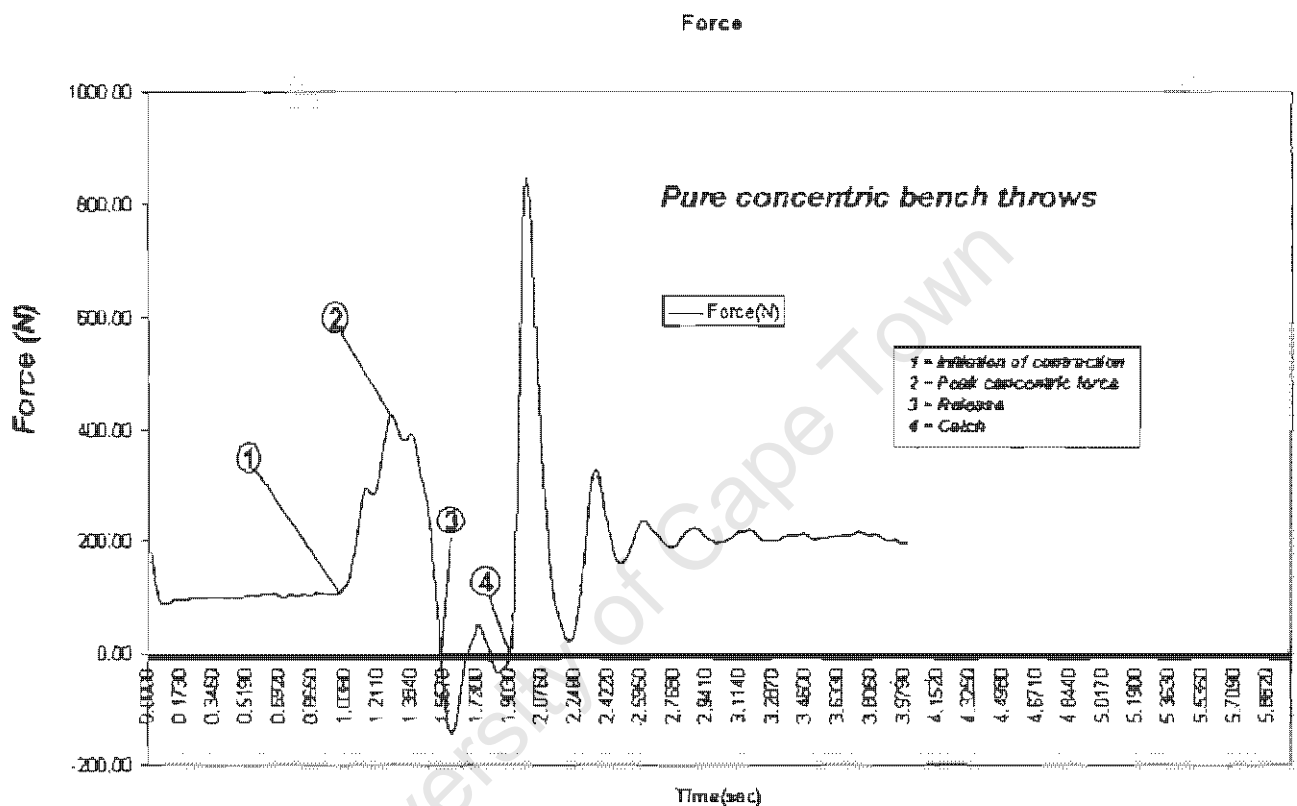
The subjects were instructed to push as fast and as hard as possible, maintaining the contraction for ~5 seconds against the isometric resistance. To ensure that the subjects gave maximal effort in the MVC, the researchers gave vocal encouragement. The maximum isometric force and corresponding integrated EMG obtained from the three trials was recorded and used to normalize the EMG values of the bench throw tests that followed. All force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

### **THE SSC TESTS FOR THE UPPER BODY**

For all the stretch shortening cycle tests, the output from the force plate was set to zero with the flat bench bolted to the top plate and the subject strapped on the bench. The sampling strategy for force data remained the same as mentioned above. For the stretch shortening cycle tests of the upper body, the pure concentric bench-throw and rebound bench-throw tests were performed. These techniques have previously been described in Chapter 2 (p 126 - 127). Pure concentric throws were performed first, followed by the rebound throws. To minimize the chances of an order effect occurring, the subjects were given as much rest as was necessary to fully recover between trials. The best of three trials for each respective throw were used for further analysis.

For the analysis of force and EMG, the data were segmented into different regions (Figures 8 and 9). For "Total" concentric measures of pure concentric bench throws (Figure 8), the data between initiation of contraction (1) and release (3) were used. For

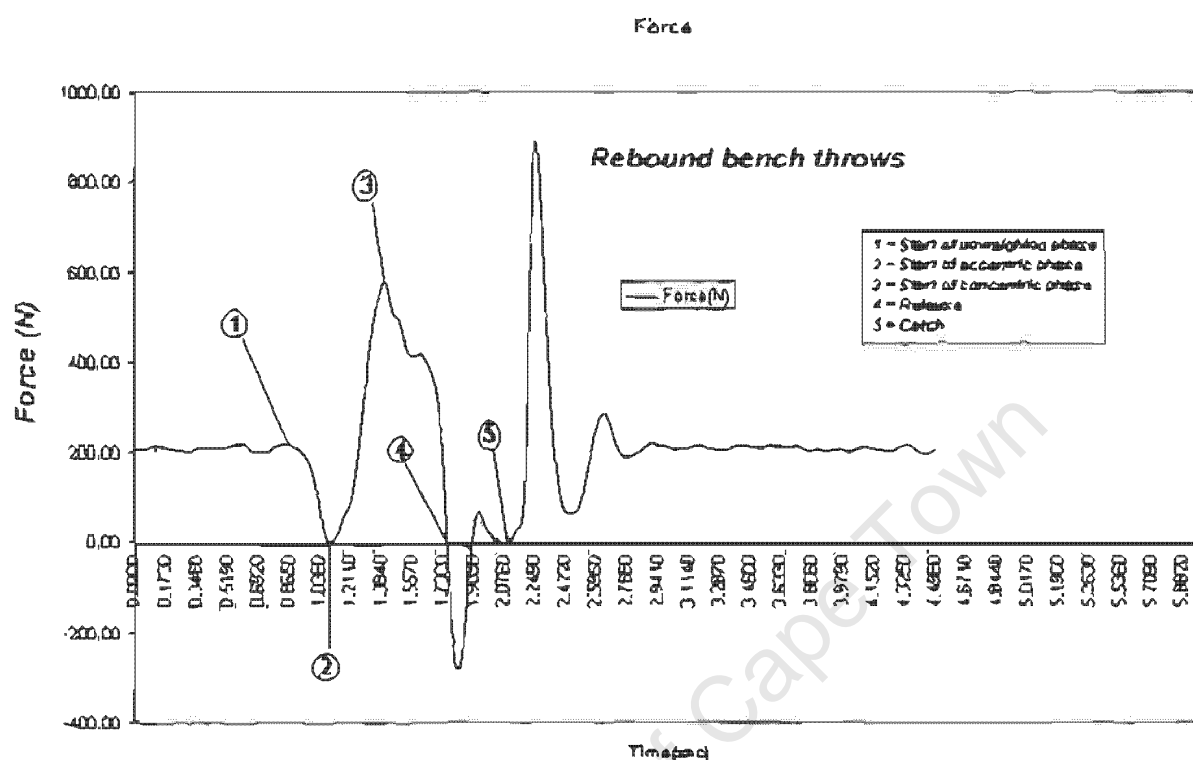
the concentric phase of pure concentric bench throws, the data between the peak concentric force (2) and release (3) were used. It was assumed that the segment between initiation of contraction (1) and peak concentric force (2) was the stretching or tensioning of the series elastic elements of the muscle-tendon complex before force transmission to the actual throw. Flight time, which is needed for the calculation of throw height, was calculated as the time difference between release (3) and catch (4).



**Figure 8:** Force-time data of a pure concentric bench throw. (\* Graph from original data)

In rebound bench throws (Figure 9), there is an un-weighting phase before eccentric contraction where the subject lowers the bar from an extended-arm position. This phase is essentially a free-falling phase and extends from the start of the un-weighting phase (1) to the start of the eccentric phase (2) where resistance to this down-falling motion is met. For "Total" stretch shortening cycle measures of rebound bench throws, the data between initiation of eccentric contraction (2) and release (4) were used. For the concentric phase of rebound bench throws, the data between the peak eccentric force, transition or start of concentric contraction (3) and release (4) were used. Flight time, which is needed for

calculation of throw height, was calculated as the time difference between release (4) and catch (5). All force and EMG data were analysed using this regional breakdown.



**Figure 9:** Force-time data of a rebound bench throw. (\* Graph from original data)

Throw height for both tests (pure concentric bench throws and rebound bench throws) was calculated using the following formulae:

- Vertical take-off velocities ( $V_v$ ) =  $\frac{1}{2} \times t_{\text{air}}(\text{flight time}) \times g$  (gravitational acceleration or  $9.81 \text{ m.s}^{-2}$ ) (7;25;27;29;101)
- Throw height ( $h$ ) =  $V_v^2/2g$  (101)
- Alternatively one could use ( $h$ ) =  $1.226 \times (t_{\text{air}})^2$  (7)

#### ELECTROMYOGRAPHY (EMG)

The anterior deltoid muscle was used as a measure of neural activation during the bench throws. Two surface EMG electrodes (Blue sensor SP-00-S, Medicotest A/S, Rugmarken, Denmark) were attached as a pair over the muscle belly with an inter-electrode distance of ~2 cm (135) and were placed so that they were orientated parallel with the active muscle fibers. A third neutral reference electrode was placed on the anterior surface of the

clavicle. Electromechanical delay was treated as a systematic error, as it was assumed that this was either negligible or constant (60).

The determination of the site for electrode placement, subject preparation, signal processing, equipment for data collection, and procedure for extraction of EMG data was the same as described for the lower body study (p 183).

STATISTICAL ANALYSIS

The data were analysed using the same statistical procedures outlined in Chapter 3 (p 135 - 137).

RESULTS

Table 15 shows the average data of the various measures used for normalization and in the calculation of stretch shortening cycle performance for each individual measured over the three trials.

Table 15: Average test data for the measures in bench throw testing used in the calculation of stretch shortening cycle performance in the upper extremities for each individual over three trials (X ± SD), (n = 10).

SUBJECT	MVC peak force (N)	MVC (kg)	MVC impulse (Ns)	MVC IEMG (mVs)	PCBT throw height (cm)	RBT throw height (cm)
SSCup1	678.5 ± 75.5	69.2 ± 7.7	1808 ± 268	1301 ± 87	17.9 ± 2.0	21.9 ± 1.5
SSCup2	518.6 ± 30.9	52.9 ± 3.2	1409 ± 89	2025 ± 1105	17.4 ± 1.6	20.1 ± 0.9
SSCup3	938.5 ± 120.3	95.7 ± 12.3	2308 ± 318	6314 ± 1122	33.2 ± 1.1	37.2 ± 1.8
SSCup4	542.8 ± 8.1	55.3 ± 0.9	1543 ± 23	2119 ± 200	16.7 ± 0.9	17.7 ± 1.4
SSCup5	1211.7 ± 28.4	123.5 ± 2.9	3073 ± 228	3426 ± 353	30.1 ± 0.5	34.8 ± 2.5
SSCup6	799.2 ± 34.6	81.5 ± 3.5	2179 ± 31	4682 ± 373	26.3 ± 1.5	28.6 ± 2.0
SSCup7	865.6 ± 72.0	91.8 ± 6.6	2555 ± 207	3818 ± 1659	28.3 ± 1.0	34.9 ± 1.6
SSCup8	626.6 ± 168.9	52.6 ± 2.9	1401 ± 85	4279 ± 1778	12.8 ± 1.4	18.8 ± 3.3
SSCup9	1501.3 ± 589.5	185.1 ± 4.7	5256 ± 169	7293 ± 789	42.3 ± 1.3	46.6 ± 2.9
SSCup10	778.3 ± 37.8	78.9 ± 3.1	2175 ± 71	4633 ± 1735	28.4 ± 1.2	30.2 ± 3.5
Mean	846.1 ± 308.1	88.6 ± 40.5	2371 ± 1145	3989 ± 1892	25.3 ± 9.1	29.1 ± 9.5

Index: MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

Table 16 shows the intra-class correlation coefficients and coefficients of variation for the various measures used for normalization and in the calculation of stretch shortening cycle performance over the three trials. Included are the 95% confidence intervals of both measures of repeatability.

Tables 15 and 16, show that most of these measures are repeatable (146). Table 16 shows that these measures have high intra-class correlation coefficients and low coefficients of variation. The only variable with a lower, yet still reasonably good repeatability, and higher variability is the MVC IEMG, which is used for normalization of EMG values in the following tests. This has a similar trend to the lower body tests presented earlier (Table 4). However, because the IEMG-values of the subsequent bench throw measures for each day of testing are normalized to the maximum IEMG or MVC IEMG, this variability was also corrected for.

**Table 16:** Repeatability of measures in stretch shortening cycle testing used in the calculation of stretch shortening cycle performance in the upper extremities ( $n = 10$ ).

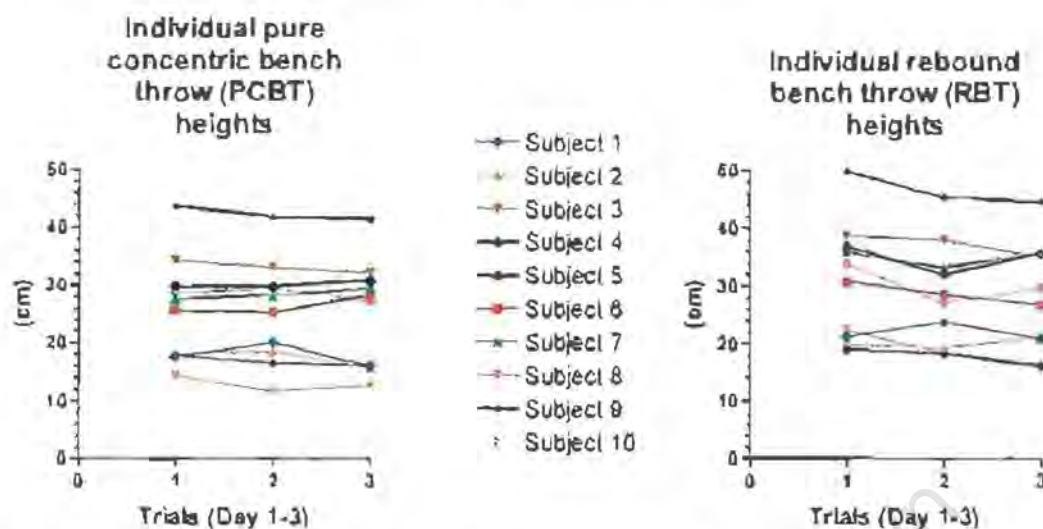
Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)
MVC Peak force (N)	0.99 (0.97-1.00)	11.7 (8.8-17.3)
MVC (kg)	0.99 (0.97-1.00)	5.7 (4.3-8.4)
MVC Impulse (Ns)	0.99 (0.97-1.00)	6.6 (5.0-9.8)
MVC IEMG (mVs)	0.89 (0.72-0.97)	24.0 (18.1-35.5)
PCBT throw height (cm)	0.99 (0.97-1.00)	5.8 (4.4-8.6)
RBT throw height (cm)	0.98 (0.94-0.99)	7.7 (5.8-11.4)
PCBT MVV (m/s)	0.99 (0.97-1.00)	2.9 (2.2-4.3)
PCBT flight time (s)	0.99 (0.97-1.00)	2.9 (2.2-4.3)
RBT MVV (m/s)	0.98 (0.94-0.99)	4.5 (3.4-6.7)
RBT flight time (s)	0.98 (0.94-0.99)	3.9 (3.0-5.8)

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw,  
RBT = rebound bench throw, MVV = maximal vertical velocity, IEMG = integrated EMG

Figure 10 shows that the throw heights, which have previously been used in research involving the upper body as indicators of stretch shortening cycle potentiation (119), are



also very repeatable between trials. As with the trial with the lower body, certain subjects however are more variable than others over the three trials.



**Figure 10:** A graphical representation of the variability for each subject of the concentric throws (PCBT) and the stretch shortening cycle throws (RBT) over the three trials.

The stretch shortening cycle throws or rebound bench throws (RBT) as with the lower body also appear to be more variable than the concentric throws or pure concentric bench throws (PCBT) (Tables 15 and 16, Figure 10).

**Table 17:** One-way analysis of variance in measures used in the calculation of stretch shortening cycle performance in the upper extremities ( $n = 10$ ).

Variable	df effect	MS effect	df error	MS error	F	p-level (*significant)
MVC Peak force (N)	2	37.37	18	3458	0.01	0.99
MVC (kg)	2	0.40	18	36.07	0.01	0.99
MVC Impulse (Ns)	2	2822	18	34976	0.08	0.92
MVC IEMG (mVs)	2	1991610	18	1149686	1.73	0.21
PCBT throw height (cm)	2	2.06	18	1.60	1.29	0.30
RBT throw height (cm)	2	20.29	18	2.86	7.10	0.01*
PCBT MVV (m/s)	2	0.01	18	0.003	1.38	0.28
PCBT flight time (s)	2	0.0001	18	0.0002	0.98	0.40
RBT MVV (m/s)	2	0.04	18	0.01	5.47	0.01*
RBT flight time (s)	2	0.002	18	0.0002	5.78	0.01*

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = Integrated EMG, MVV = Maximal vertical velocity

The one-way ANOVA showed that there were significant differences in the rebound bench throw heights and their determining parameters i.e. maximal vertical velocity and flight times over the three trials (Table 17).

A Scheffe's post-hoc test was performed on these data. Table 18 shows that the rebound bench throws (and their determining variables), which are representative of stretch shortening cycle ability in the upper body, decreased significantly over the three testing days.

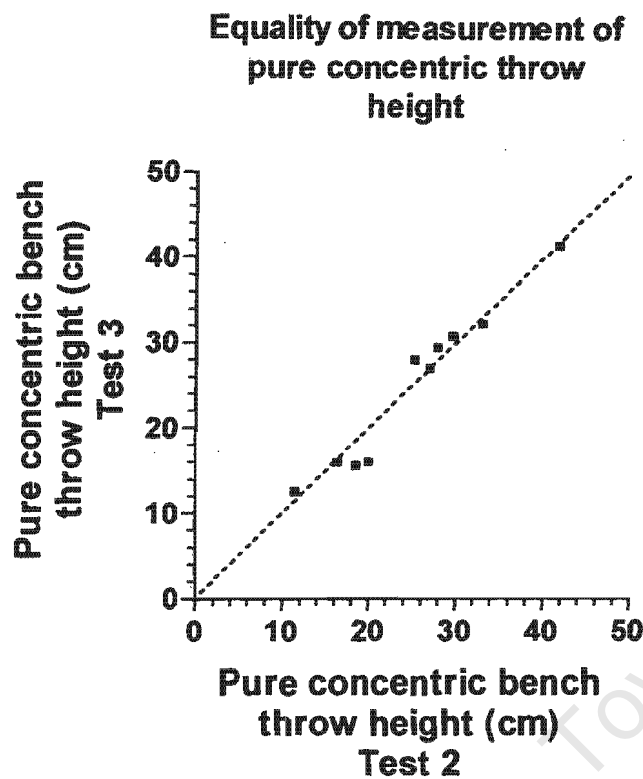
**Table 18:** Scheffe's Post-hoc test for differences between trials in the upper body stretch shortening cycle tests ( $n = 10$ ).

Variable: RBT (*significant)			
Testing day	Average day 1: 30.8 cm	Average day 2: 28.57	Average day 3: 28.15
1		0.03*	0.01*
2	0.03*		0.86
3	0.01*	0.86	
Variable: RBT MVV (*significant)			
Testing day	Average day 1: 2.43 m.s	Average day 2: 2.33 m.s	Average day 3: 2.30 m.s
1		0.08	0.02*
2	0.08		0.78
3	0.02*	0.78	
Variable: RBT FLT (*significant)			
Testing day	Average day 1: 0.50 s	Average day 2: 0.48 s	Average day 3: 0.47 s
1		0.06	0.02*
2	0.06		0.80
3	0.02*	0.80	

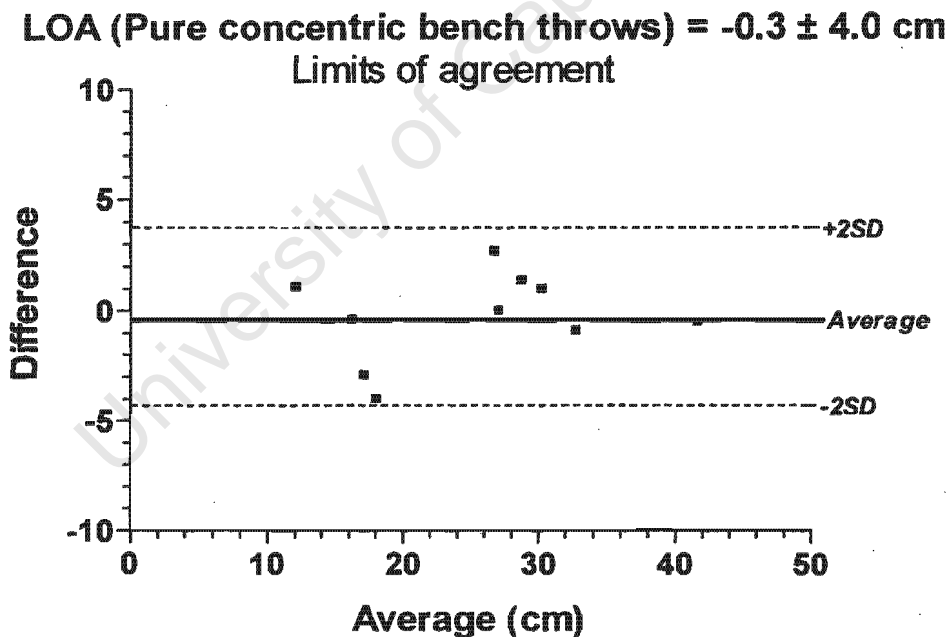
**Index:** RBT = rebound bench throw, MVV = maximal vertical velocity, FLT = flight time

Figure 11 indicates that the precision of the measurement of pure concentric bench throw height, as determined by the limits of agreement (two standard deviations above and below the average difference between trials) using the NAMS unit, falls within a range of 8.0 cm ( $-0.3 \pm 4.0$  cm) (16).

**A**



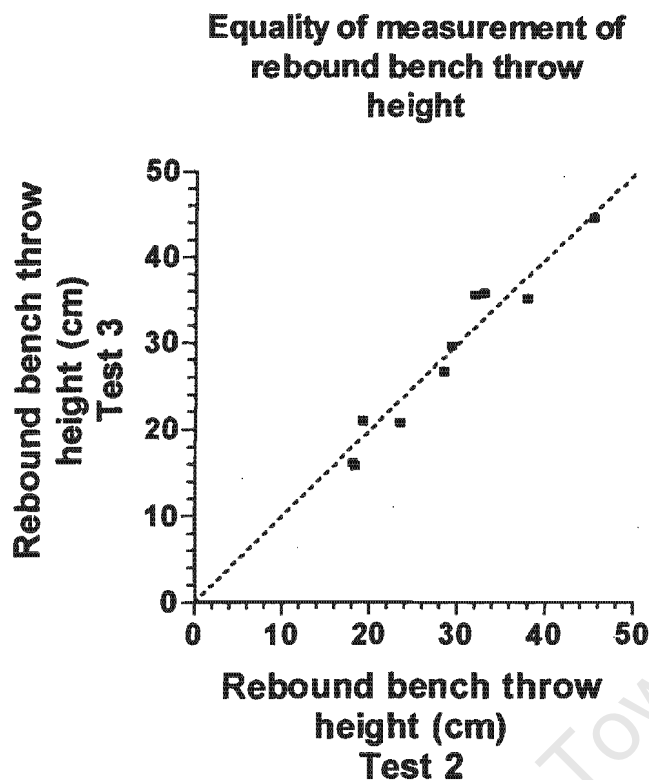
**B**



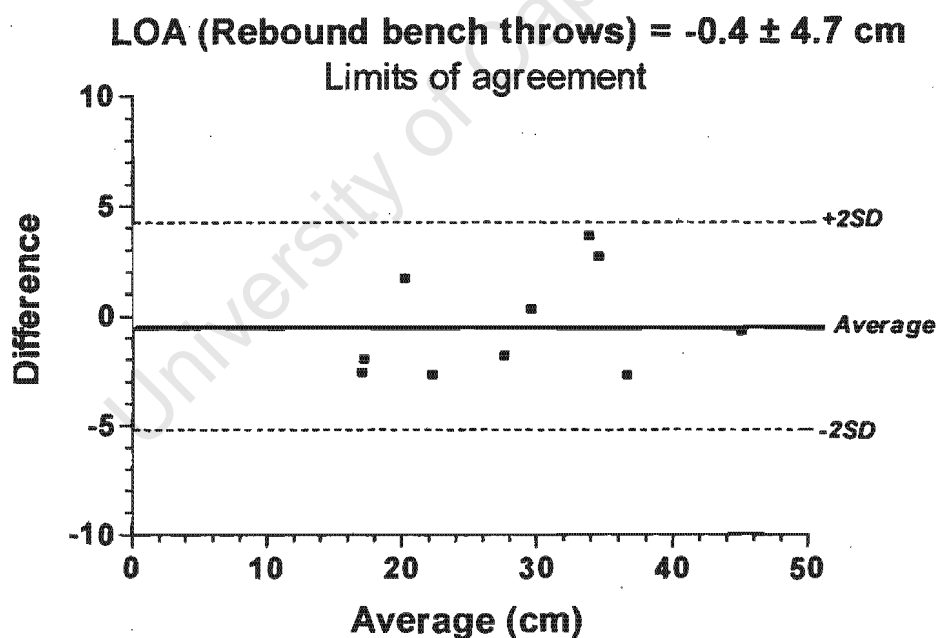
**Figure 11:** A - The equality of measurement for pure concentric bench throw height between the second and third testing day using the NAMS unit. B - The limits of agreement for measurement of pure concentric bench throw height between the second and third testing day using the method of Bland-Altman (16).

Figure 12 indicates that the precision of the measurement of rebound bench throw height, as determined by the limits of agreement (two standard deviations above and below the average difference between trials) using the NAMS unit, falls within a range of 9.4 cm ( $-0.4 \pm 4.7$  cm) (16).

**A**



**B**



**Figure 12:** A - The equality of measurement for rebound bench throw height between the second and third testing day using the NAMS unit. B – The limits of agreement for measurement of rebound bench throw height between the second and third testing day using the method of Bland-Altman (16).

Table 19 shows the average data for each subject for the "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials.

**Table 19:** Average test data for "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities (n = 10).

SUBJECT	"Total" PCBT impulse (Ns)	"Total" PCBT IEMG (%MVC)	"Total" RBT impulse (Ns)	"Total" RBT IEMG (%MVC)
SSCup1	171.4 ± 7.3	22.4 ± 3.6	216.0 ± 5.9	22.9 ± 5.8
SSCup2	164.5 ± 10.7	18.0 ± 4.6	227.4 ± 9.7	27.2 ± 5.6
SSCup3	202.4 ± 5.3	15.2 ± 2.5	274.7 ± 6.0	19.8 ± 1.0
SSCup4	155.4 ± 13.0	22.2 ± 2.1	206.4 ± 3.3	20.6 ± 3.6
SSCup5	158.8 ± 5.7	10.6 ± 1.4	222.8 ± 6.3	12.6 ± 1.9
SSCup6	180.9 ± 18.0	14.8 ± 1.8	233.7 ± 10.1	17.6 ± 3.8
SSCup7	157.5 ± 7.9	10.9 ± 3.6	226.0 ± 9.7	17.2 ± 6.0
SSCup8	153.3 ± 9.0	19.9 ± 3.4	236.8 ± 18.1	22.7 ± 5.3
SSCup9	176.2 ± 12.7	7.9 ± 1.2	245.7 ± 2.3	7.4 ± 3.6
SSCup10	155.0 ± 0.9	10.9 ± 1.2	222.1 ± 5.9	14.6 ± 1.3
Mean	167.6 ± 15.5	15.3 ± 5.2	231.2 ± 18.8	18.3 ± 5.7

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

Table 20 shows the intra-class correlation coefficients and coefficients of variation for the "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials. Included are the 95% confidence intervals of the both measures of repeatability.

**Table 20:** Repeatability of "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities (n = 10).

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	% Coefficient of Variation (95% Confidence Interval)
"Total" PCBT impulse (Ns)	0.86 (0.65-0.96)	5.4 (4.1-8.0)
"Total" PCBT IEMG (%MVC)	0.90 (0.74-0.97)	16.9 (12.8-25.0)
"Total" RBT impulse (Ns)	0.93 (0.81-0.98)	3.3 (2.5-4.9)
"Total" RBT IEMG (%MVC)	0.82 (0.57-0.95)	22.0 (16.6-32.5)

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

The "Total" force-related i.e. impulse measures also have high repeatability (Table 20). However the EMG-associated measures especially with the rebound bench throws were less repeatable (R = 0.82 (95% CI: 0.57-0.95); CV = 22.0 (95% CI: 16.6-32.5)). As with the lower body tests, it seems as if the EMG related variables have the highest variation. However, the variation of these EMG-related measures is still within acceptable limits (146).

The one-way ANOVA showed that there were no significant differences over the three trials between the "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials (Table 21).

**Table 21:** One-way analysis of variance in "Total" force and -EMG measures used in stretch shortening cycle testing of the upper extremities (n = 10).

Variable	df effect	MS effect	df error	MS error	F	p-level (*significant)
"Total" PCBT impulse (Ns)	2	26.39	18	111.11	0.24	0.79
"Total" PCBT IEMG (%MVC)	2	16.66	18	6.71	2.48	0.11
"Total" RBT impulse (Ns)	2	68.49	18	79.00	0.87	0.44
"Total" RBT IEMG (%MVC)	2	28.29	18	16.29	1.74	0.20

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

Table 22 shows the average data for each subject for the concentric force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials.

**Table 22:** Average test data for concentric force and -EMG measures in stretch shortening cycle testing of the upper extremities (n = 10).

SUBJECT	PCBT concentric impulse (Ns)	PCBT concentric IEMG (%MVC)	RBT concentric impulse (Ns)	RBT concentric IEMG (%MVC)
SSCup1	84.3 ± 10.6	10.1 ± 2.7	130.9 ± 4.5	15.3 ± 4.5
SSCup2	92.9 ± 3.6	10.7 ± 2.2	145.8 ± 14.5	17.3 ± 2.2
SSCup3	87.7 ± 17.7	4.3 ± 0.2	111.4 ± 13.5	5.8 ± 1.3
SSCup4	77.1 ± 4.0	10.5 ± 1.6	130.0 ± 7.2	14.3 ± 3.9
SSCup5	88.2 ± 23.9	5.9 ± 2.3	132.7 ± 2.2	7.1 ± 0.3
SSCup6	81.3 ± 18.1	5.5 ± 1.3	131.6 ± 16.0	9.8 ± 3.6
SSCup7	79.3 ± 6.2	5.5 ± 1.9	131.4 ± 5.9	10.1 ± 3.2
SSCup8	77.3 ± 6.0	8.2 ± 1.6	125.5 ± 8.1	13.5 ± 2.3
SSCup9	78.2 ± 21.5	3.2 ± 1.2	134.9 ± 11.6	5.0 ± 2.5
SSCup10	74.5 ± 3.7	5.2 ± 1.8	125.4 ± 7.3	8.1 ± 0.8
Mean	82.1 ± 5.9	6.9 ± 2.7	130.0 ± 8.6	10.6 ± 4.3

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

Table 23 shows the intra-class correlation coefficients and coefficients of variation for the concentric force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials. Included are the 95% confidence intervals of both the measures of repeatability.

**Table 23:** Repeatability of concentric force and -EMG measures in stretch shortening cycle testing of the upper extremities ( $n = 10$ ).

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	% Coefficient of Variation (95% Confidence Interval)
PCBT concentric impulse (Ns)	-0.79 ((-0.59)- (-2.75))	13.9 (10.5-20.6)
PCBT concentric IEMG (%MVC)	0.85 (0.63-0.96)	25.7 (19.4-38.0)
RBT concentric impulse (Ns)	0.55 (0.17-0. 84)	7.0 (5.3-10.4)
RBT concentric IEMG (%MVC)	0.86 (0.65-0.96)	24.1 (18.2-35.6)

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG

It is noticeable that the concentric measurements in Tables 22 and 23, as with the lower body (Tables 9 and 10), have a higher variability and lesser repeatability than the previous "Total" data tables (Tables 15 and 20). This is evident in the greater confidence interval ranges of intra-class correlation coefficients and coefficients of variation. The one-way ANOVA showed that there were no significant differences in the concentric force and -EMG measures used in stretch shortening cycle testing of the upper extremities over the three trials (Table 24).

**Table 24:** One-way analysis of variance in concentric force and -EMG measures in stretch shortening cycle testing of the upper extremities ( $n = 10$ ).

Variable	Df effect	MS effect	df error	MS error	F	p-level (*significant)
PCBT concentric impulse (Ns)	2	294.74	18	178.77	1.65	0.22
PCBT concentric IEMG (%MVC)	2	2.09	18	3.42	0.61	0.55
RBT concentric impulse (Ns)	2	65.68	18	105.11	0.62	0.55
RBT concentric IEMG (%MVC)	2	12.86	18	7.11	1.81	0.19

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, IEMG = integrated EMG



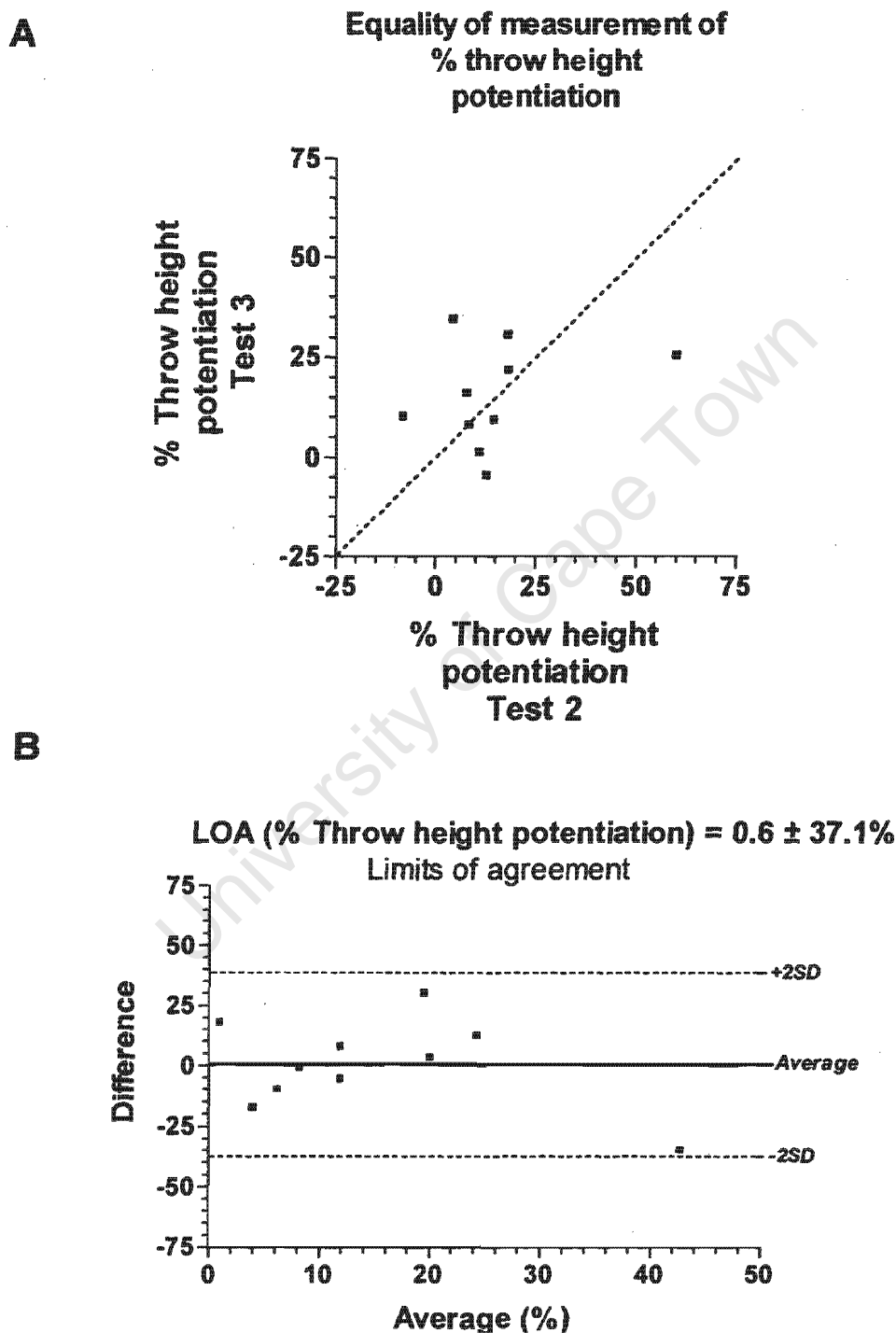
Table 25 shows the intra-class correlation coefficients, coefficients of variation and respective 95% confidence intervals of all the possible mathematical derivatives that could be used to calculate or measure stretch shortening cycle potentiation of performance in the upper body musculature. Table 25 also includes the results of the one-way ANOVA for significant difference between trials, with significance accepted as  $p < 0.05$ .

**Table 25:** Repeatability of mathematical derivatives of stretch shortening cycle performance in stretch shortening cycle testing of the upper extremities ( $n = 10$ ).

Variable	Intra-class Correlation Coefficient (95% Confidence Interval)	%Coefficient of Variation (95% Confidence Interval)	p-level (*) significant difference between trials
"Total" PCBT F:EMG ratio (N%/MVC)	0.90 (0.74-0.97)	17.6 (13.3-26.0)	0.49
"Total" RBT F:EMG ratio (N%/MVC)	0.72 (0.40-0.91)	23.1 (17.5-34.2)	0.35
Concentric PCBT F:EMG ratio (N%/MVC)	0.88 (0.70-0.97)	22.7 (17.2-33.6)	0.10
Concentric RBT F:EMG ratio (N%/MVC)	0.74 (0.43-0.92)	25.3 (19.1-37.4)	0.40
"Total" RBT/"Total" PCBT Impulse potentiation ratio	0.50 (0.11-0.82)	5.9 (4.5-8.7)	0.92
RBT/PCBT concentric Impulse potentiation ratio	0.08 (-0.23-0.55)	11.7 (8.8-17.3)	0.24
RBT-PCBT Height difference (cm)	0.48 (0.09-0.81)	71.5 (54.0-105.7)	0.07
RBT-PCBT height % Potentiation	0.75 (0.45-0.92)	71.7 (54.2-106.0)	0.36
RBT/PCBT height potentiation ratio	0.77 (0.48-0.93)	7.0 (5.3-10.4)	0.40
"Total" RBT-PCBT F:EMG difference (N%/MVC)	0.42 (0.03-0.78)	162.3 (122.6-240.0)	0.46
"Total" RBT-PCBT F:EMG % difference	0.47 (0.08-0.81)	129.1 (97.6-190.9)	0.63
"Total" RBT F:EMG/"Total" PCBT F:EMG ratio	0.47 (0.08-0.81)	17.7 (13.4-26.2)	0.63
RBT-PCBT concentric F:EMG difference (N%/MVC)	-0.09 (-0.33-0.37)	743.8 (562.0-1100.0)	0.87
RBT-PCBT concentric F:EMG % difference	-0.24 (-0.40-0.16)	733.5 (554.2-1084.7)	0.39
Concentric RBT F:EMG/ Concentric PCBT F:EMG ratio	-0.24 (-0.40- -0.16)	20.6 (15.6-30.5)	0.39

**Index:** MVC = maximal voluntary contraction, PCBT = pure concentric bench throw, RBT = rebound bench throw, EMG = integrated EMG, F = integrated force

Figure 13 indicates that the precision (two standard deviations above and below the average difference between trials) by limits of agreement according to Bland-Altman (16) for measurement of throw height potentiation using the NAMS unit falls within a range of 74.2 % ( $0.6 \pm 37.1$  %).



**Figure 13:** A - The equality of measurement for throw height potentiation via the stretch shortening cycle between the second and third testing day using the NAMS unit. B – The limits of agreement for measurement of throw height potentiation between the second and third testing day using the method of Bland-Altman (16).

From the Table 25, it can clearly be seen that the most repeatable derivative measures are those of the "Total" PCBT F: EMG-, "Total" RBT F: EMG-, Concentric PCBT F: EMG- and Concentric RBT F: EMG ratios. It would seem as if the pure concentric bench throw data were generally more repeatable than the rebound bench throw data. The remainder of the mathematical derivatives all lack significant repeatability. Of great importance is the fact that the main measure of stretch shortening cycle potentiation i.e. RBT-PCBT % height potentiation was not very repeatable and was variable. This is also visible in Figure 13, where the precision of measure (16) is also indicated. However, when expressing the exact same data in a different way i.e. RBT/PCBT height potentiation ratio, the variability of the measure improves significantly. This raises some serious questions about the effective use of mathematical derivatives in calculating potentiation.

## DISCUSSION

The main aim of this study was to test, using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town), whether the various measures that are used in testing for the stretch shortening cycle performance of the upper body musculature were repeatable.

The first main finding of this study is that the bench throw height measures using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town) were repeatable over three different testing days (Table 16).

The static or pure concentric bench throws showed an intra-class correlation coefficient of 0.99 (95% CI: 0.97-1.00) and coefficient of variation of 5.8 (95% CI: 4.4-8.6)%, which is better than Wilson et al. (155), who measured an intra-class correlation coefficient of  $R = 0.85$  using a similar test to the pure concentric bench throw. The stretch shortening cycle or rebound bench throws showed an intra-class correlation coefficient of 0.98 (95% CI: 0.94-0.99) and coefficient of variation of 7.7 (95% CI: 5.8-11.4)%. Although both measures (Table 16) were repeatable, as with the lower body (Table 3), the stretch shortening cycle or rebound throws seemed to be more variable than the pure concentric bench throws. This conclusion is supported by and further visible when one examines Figure 10, which compares the trends of each individual in both, throws, over the three days.

Some subjects also needed more time to become familiarized with the rebound bench throws, which also occurred with the countermovement jumps of the lower body. There was less learning effect with the pure concentric movements in the upper (PCBT) and lower body (SJ).

Even though the bench throw height data has been shown to be quite repeatable, one should also take note of the range of precision of measurement for future testing. Determined by the criteria of Bland-Altman (16), the precision of measurement of the throw height data, for pure concentric bench throws falls within a range of 8.0 cm and for rebound bench throws falls within a range of 9.4 cm. So for any differences to be measured in future research, this needs to be taken into consideration (Figures 11 and 12).

The second finding that is supported by Figure 10 and Table 18 is that the stretch shortening cycle ability of the upper body musculature might be affected by fatigue. The one-way ANOVA in Table 17 shows that, even though they were repeatable measures (Table 16), there was a significant difference between the rebound throws and their determining parameters i.e. maximal vertical velocity and flight time, for the group over the three days.

The Scheffe's post-hoc test that followed (Table 18) showed that there was a significant, progressive decline in rebound bench throw height and its determining parameters over the three days. Even though there was 48 hours rest in between testing days, it seems as if accumulative fatigue over the four testing days (familiarization included) might have had an influence on this measure of stretch shortening cycle performance and could possibly account for the increased variability of the rebound bench throws versus the pure concentric bench throws. This needs to be taken into account when experiments are designed.

The third finding in this study was that the measures of effort used for normalization (Table 16) i.e. MVC peak force (N) ( $R = 0.99$  (95% CI: 0.97-1.00);  $CV = 11.7$  (95% CI: 8.8-17.3)%), MVC peak load (kg) ( $R = 0.99$  (95% CI: 0.97-1.00);  $CV = 5.7$  (95% CI: 4.3-8.4)% and MVC impulse ( $R = 0.99$  (95% CI: 0.97-1.00);  $CV = 6.6$  (95% CI: 5.0-9.8)%)

as a control measure of maximal effort, were highly repeatable over the three trials. These data show, as with the lower body, that the subjects exerted maximal efforts for each day.

The MVC IEMG ( $R = 0.89$  (95% CI: 0.72-0.97);  $CV = 24.0$  (95% CI: 18.0-35.5)%) were still within acceptable limits, however, varied substantially more than the force-time measures. This was also found in the lower body measures. Although more variable, the subsequent EMG data were normalized against the MVC IEMG measured on each respective day, which corrected for the variability in this measure. The increased variability could be due to various reasons as mentioned earlier e.g. minor positional differences in electrode placement, electrical conductivity of the skin surface, subcutaneous fat, etc. .

The fourth finding of this study was that the *"Total"* force and -IEMG measures (Table 20) had a similar pattern of repeatability compared to the stretch shortening cycle performance measures (Table 16). The force-time measures i.e. *"Total"* PCBT impulse ( $R = 0.86$  (95% CI: 0.65-0.96);  $CV = 5.4$  (95% CI: 4.1-8.0)) and *"Total"* RBT impulse ( $R = 0.93$  (95% CI: 0.81-0.98);  $CV = 3.3$  (95% CI: 2.5-4.9)) were repeatable. In this instance the pure concentric measure (PCBT impulse) was less repeatable. The EMG measures i.e. *"Total"* PCBT IEMG ( $R = 0.90$  (95% CI: 0.74-0.97);  $CV = 16.9$  (95% CI: 12.8-25.0)) and *"Total"* RBT IEMG ( $R = 0.82$  (95% CI: 0.57-0.95);  $CV = 22.0$  (95% CI: 16.6-32.5)), were also repeatable, but they were however far more variable than the force-time measures. This could possibly be due to different recruitment patterns during the throws, or as mentioned earlier, minor electrode positional differences, subcutaneous fat, and skin surface electrical conductivity.

The fifth finding in this study was that the *"Concentric-defined"* measures (Table 23) were generally not repeatable. The *"Concentric-defined"* IEMG measures i.e. PCBT concentric IEMG ( $R = 0.85$  (95% CI: 0.63-0.96);  $CV = 25.7$  (95% CI: 19.4-38.0)) and RBT concentric IEMG ( $R = 0.86$  (95% CI: 0.65-0.96);  $CV = 24.1$  (95% CI: 18.2-35.6)) were similar in repeatability to the *"Total"* IEMG measures (Table 20). The force-time measures PCBT concentric impulse ( $R = -0.79$  (95% CI: (-0.59)-(-2.75));  $CV = 13.9$  (95% CI: 10.5-20.6)) and RBT concentric impulse ( $R = 0.55$  (95% CI: 0.17-0.84);  $CV = 7.0$  (95% CI: 5.3-10.4)) indicated low repeatability and therefore should not be used in further studies.

The sixth important finding of this study was that most of the mathematical derivatives (Table 25), which are often used to calculate or measure stretch shortening cycle performance or potentiation in the upper body musculature, were either not repeatable, or highly variable. A similar result was found in the lower body study. Stretch shortening cycle performance of the muscle-tendon complex measured as the height difference between pure concentric bench throws and rebound bench throws has been used in previous studies using a similar protocol (119). As found in the lower body data, we raise some serious doubts as to the ability of the % height potentiation index and other index variations with the same data to accurately gauge stretch shortening cycle performance or stretch shortening cycle potentiation. Even though the individual measures of pure concentric bench throws ( $R = 0.99$  (95% CI: 0.97-1.00);  $CV = 5.8$  (95% CI: 4.4-8.6)) and rebound bench throws ( $R = 0.98$  (95% CI: 0.94-0.99);  $CV = 7.7$  (95% CI: 5.8-11.4)) were highly repeatable (Table 16), the mathematical derivative (Table 25, Figure 13) i.e. % bench throw height potentiation ( $R = 0.75$  (95% CI: 0.45-0.92);  $CV = 71.7$  (95% CI: 54.2-106.0)) was not.

## CONCLUSION

The main conclusion of this study is that the measurement of stretch shortening cycle performance of the upper body is generally repeatable where measured with the NAMS unit (Zest Manufacturing, PTY (Ltd) and University of Cape Town). However, the "Concentric-defined" measurements especially in the upper body, lack sufficient repeatability and have a higher variability and therefore should be used with caution. One should also consider that the EMG-related measures are generally more variable and less repeatable than the force-time measures.

As with the lower body data, many of the actual data or individual measurements of stretch shortening cycle performance are repeatable. However, when these are incorporated into ratios and mathematical derivatives, they are no longer necessarily repeatable or reliable (Table 25).

## SUMMARY OF FINDINGS

The jump height and bench throw height measures using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town) are repeatable and have sufficient precision to detect changes with about 10 subjects. However, when familiarizing subjects with these techniques more attention needs to be applied to the stretch shortening cycle tests i.e. countermovement jumps and rebound bench throws. These tests are more difficult to learn, and to ensure a better reliability of the measure it would be prudent to take this into account.

Additionally with the rebound bench throws, more rest between repetitions and tests needs to be given to prevent fatigue from influencing the rebound throw height and thereby improving the accuracy of the measure. It might also be prudent to use this same approach in the lower body test.

The MVC peak force, MVC peak load and MVC maximal effort measures of both the upper and lower body tests using the NAMS unit (Zest Manufacturing PTY (Ltd) and University of Cape Town) are all also very repeatable.

The "Total" force and "Total" IEMG measures of both upper and lower body tests are also repeatable. However, the "Concentric-defined" force and IEMG measures of both upper and lower body measures are not as repeatable and should not be used with a relatively small sample size ( $\pm 10$  subjects).

One should also be cautious in the interpretation of stretch shortening cycle potentiation measures at present. The mathematical manipulation of very repeatable data sets e.g. squat jump height and countermovement jump height in the lower body, and pure concentric bench throw height and rebound bench throw height in the upper body, seems to be flawed when combined and expressed as differences, ratios or relative change indices, as this reduces the reliability and increases the variability of these derived measures. To represent the potentiation in muscle performance via the stretch shortening cycle, one needs to somehow relate the pure concentric measures with the stretch shortening cycle measures, and at present there does not seem to be a better alternative



in quantifying this relationship. One should however, in future research, acknowledge and understand the limitations of the derived measures of potentiation expressed in this way.

University of Cape Town

## CHAPTER 6:

# What is the relationship between tendon stiffness and stretch shortening cycle performance in the lower body?

## INTRODUCTION

The next phase of this thesis was to relate maximal musculotendinous (tendon) stiffness (Chapter 3) and stretch shortening cycle ability of the lower body (Chapter 5) using the NAMS unit, to answer the following research question: *"Is there a relationship between musculotendinous stiffness and stretch shortening cycle muscle function in the lower body musculature?"*

Although previous research has shown that tendon stiffness has a significant impact on performance improvement during stretch shortening cycle activities when compared to isolated concentric activities in the same muscle groups (95;136;149;153;155), questions were raised about the accuracy of these interpretations (Chapters 3 and 5).

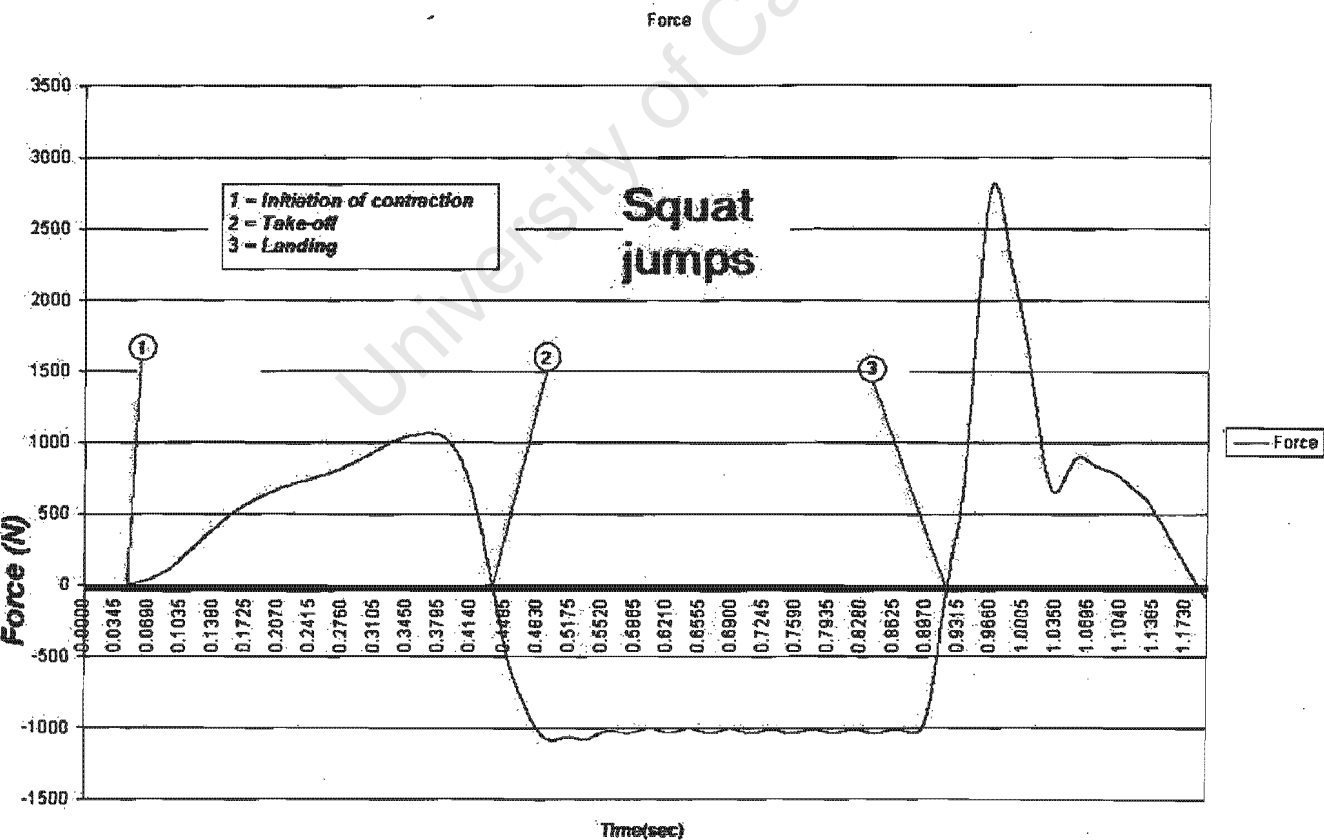
In Chapter 3 the relationships between the strength and body mass of the subjects and their maximal (tendon) stiffness values were considered. Some additional methodological concerns were also raised. As a result the technique was modified (Chapter 4) to use an absolute incremental-loading protocol rather than a relative incremental protocol. In addition, an alternative equation was used to fit the stiffness vs. load data. This modification to the protocol improved the measurement of the stiffness-load characteristics of muscle and tendon.

In Chapter 5, it was evident that the measures used to describe potentiation after stretch shortening cycle activity e.g. percentage jump height improvement, as they have been described in previous studies (73;97;101;151;152) lack repeatability, even though their individual measures e.g. squat jump height and countermovement jump height, were repeatable. This raised the question of the most appropriate way to accurately measure stretch shortening cycle potentiation.

In attempt to answer this question, we analysed the testing procedure, given the experience gained in the study described in Chapter 5. Based on logical reasoning, the following changes were made to the testing protocol.

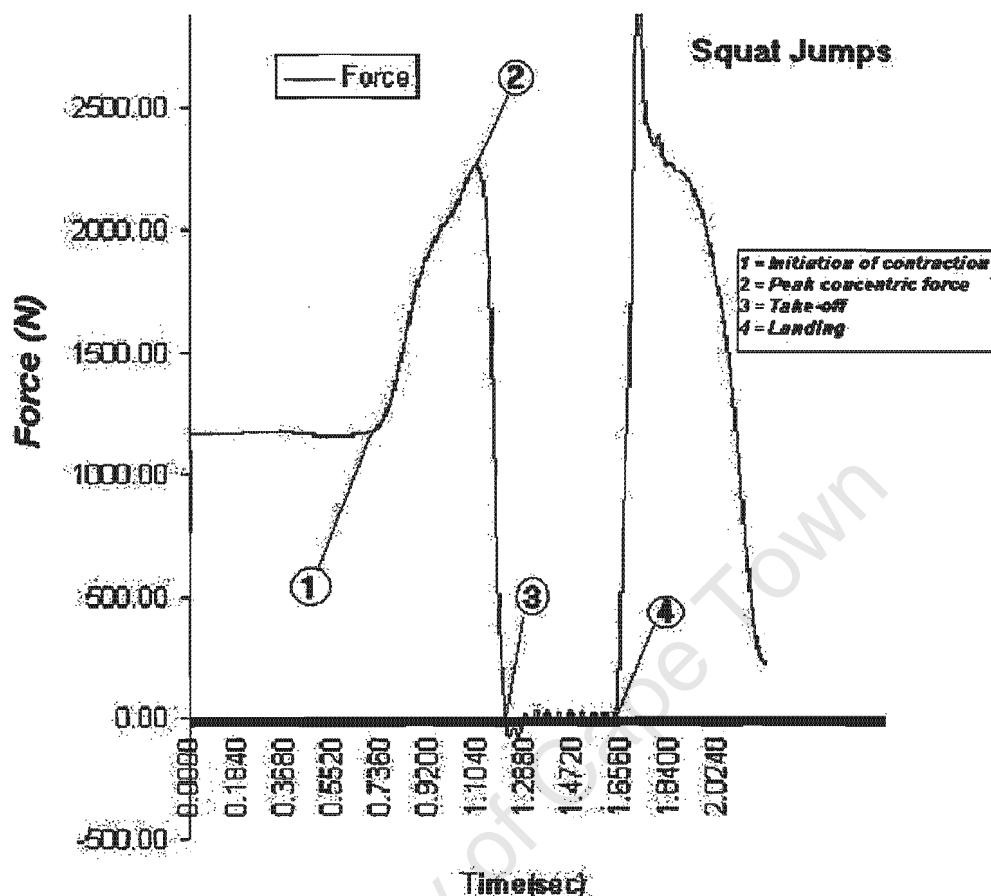
In the initial testing procedure (Chapter 5), we did not normalize for the body- and bar weight when testing. The data gathered were nevertheless repeatable (see Chapter 5). However, human judgment was often involved in determining the initiation of contraction, sometimes the take-off and sometimes the landing on the force plate, where vibration of the top plate hindered the accuracy of the measurement. It is logical to assume that by eliminating human error the repeatability of these measures would improve resulting in greater accuracy of the potentiation measures or mathematical derivatives thereof.

Therefore the protocol was adapted by zeroing the force-measurement of the force-plate to accommodate the weight of the subject and bar, thereby normalising each individual for the combined subject-bar weight (Figure 1).



**Figure 1:** Modified force-time regional breakdown of a squat jump after normalising the force-plate to accommodate the weight of the subject and bar. (\* Graph from original data)

Figure 2 shows the original regional breakdown of squat jumps used in Chapter 5. These data were not normalised for the combined bar-subject weight.



**Figure 2:** Original force-time regional breakdown of a squat jump (Chapter 5). (\* Graph from original data)

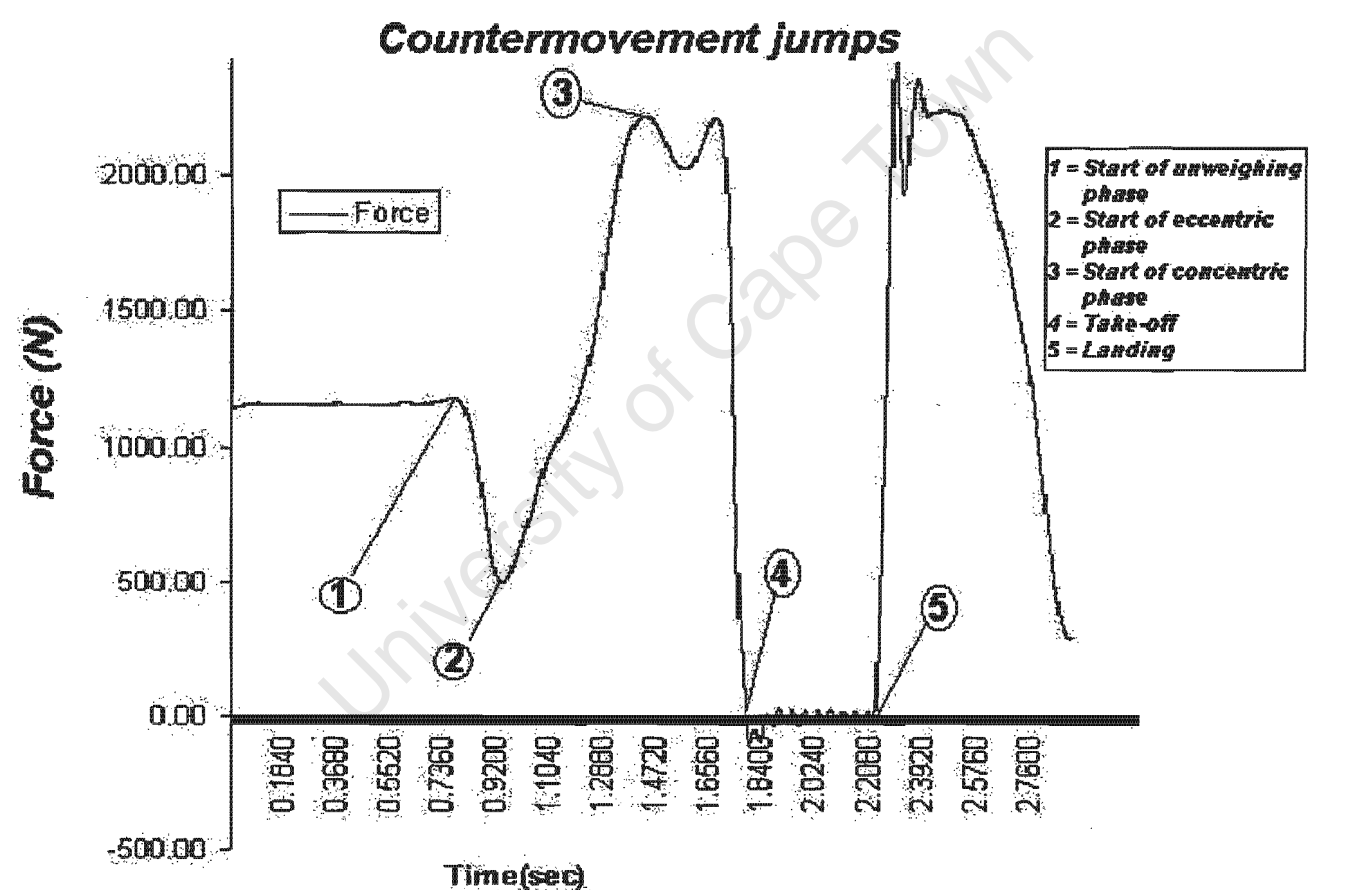
A new regional breakdown was defined when using the normalised force-time data. As the “Total” measures of Chapter 5 were generally the most repeatable measurements and the “Concentric-defined” measures were generally not, the “Concentric-defined” measures were excluded from the modified regional breakdown (Figure 1).

By zeroing the force-plate and normalising the data for the combined bar-subject weight, the determination of the force-time regional breakdown would be less subjective and more objective, thereby reducing experimental error (Figure 1). On this premise, all the original study jump height data were recalculated (Chapter 5) with the modified normalised breakdown to compare the repeatability results. The take-off (2) and landing (3) points (Figure 1) were adjusted to represent when the bar-subject weight started to leave the force plate and when the full bar-subject weight had returned to the force-plate

respectively. This adjustment was treated as a systematic error as it was assumed that it was constant.

Squat jump height originally had an intra-class correlation coefficient (R) of 0.98 (95% CI: 0.94-0.99) and coefficient of variation (CV) of 3.6 (95% CI: 2.7-5.3)%. The normalised breakdown improved the repeatability of the squat jumps slightly to R = 0.98 (95% CI: 0.94-1.00) and CV = 2.9 (95% CI: 2.2-4.4)%.

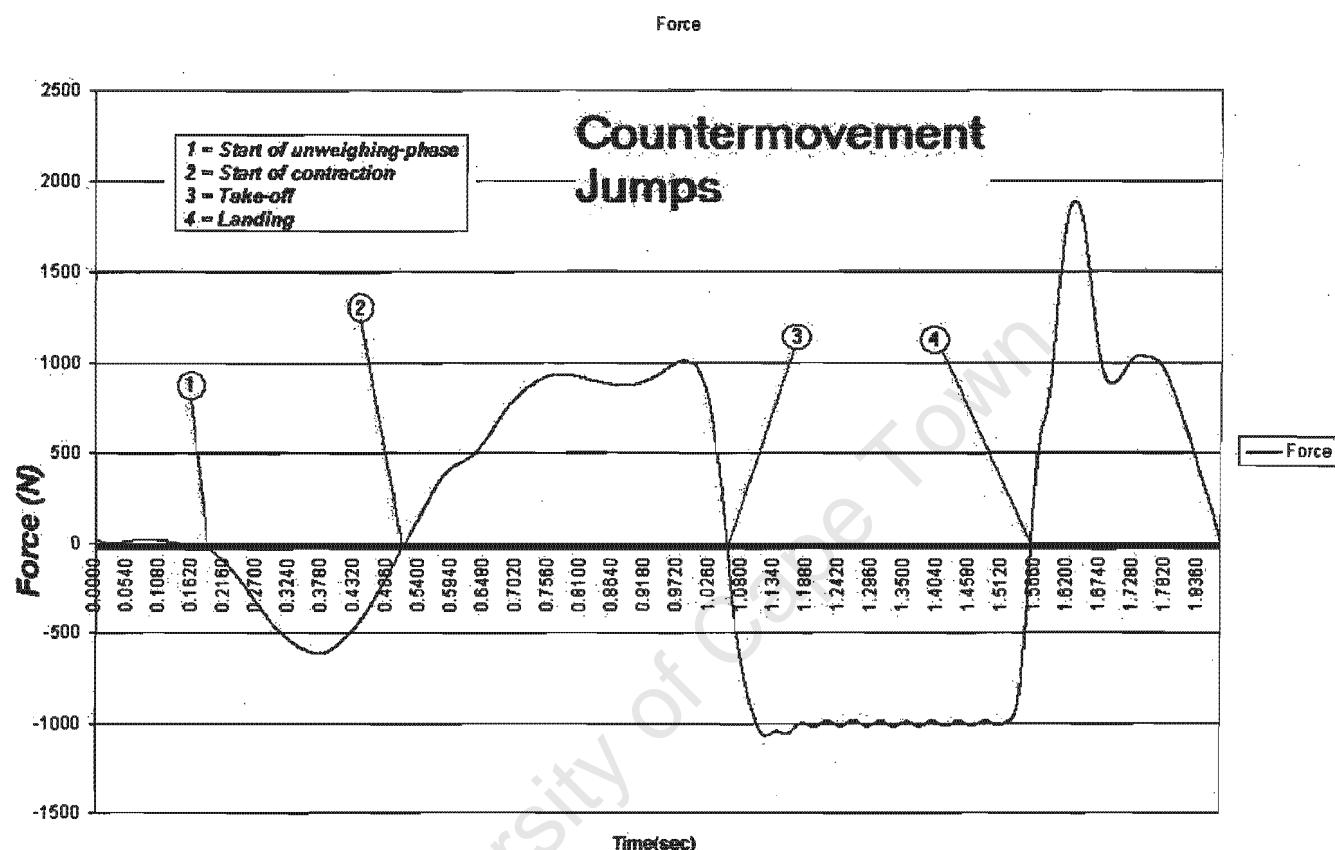
Figure 3 shows the original regional breakdown of countermovement jumps used in Chapter 5. These data were also not normalised for the combined bar-subject weight.



**Figure 3:** Original force-time regional breakdown of a countermovement jump (Chapter 5).  
(\* Graph from original data)

As with the squat jumps, the “Total” measures of Chapter 5 were the most repeatable and the “Concentric-defined” measures were also excluded from the modified regional breakdown (Figure 4). The take-off (3) and landing (4) points (Figure 4) as with the squat jumps, were adjusted to represent when the bar-subject weight started to leave the force

plate and when the full bar-subject weight had returned to the force-plate respectively. Initiation of contraction (2) was also adjusted to when the deceleration- or eccentric force equalled that of the combined bar-subject weight. These adjustments were also treated as a systematic error as it was assumed that it was constant.



**Figure 4:** Modified force-time regional breakdown of a countermovement jump after normalising the force-plate to accommodate the weight of the subject and bar. (\* Graph from original data)

A similar trend to the squat jumps was found with the countermovement jumps. Countermovement jump height originally had an intra-class correlation coefficient  $R = 0.96$  (95% CI: 0.89-0.99) and coefficient of variation  $CV = 6.4$  (95% CI: 4.8-9.5)%. The normalised breakdown improved the repeatability of the countermovement jumps to  $R = 0.98$  (95% CI: 0.94-1.00) and  $CV = 3.2$  (95% CI: 2.4-4.9)%.

This improved repeatability also reflected in an improved precision (two standard deviations above and below the average difference between trials) as calculated by the limits of agreement (16). The limits of agreement of the squat jumps improved from a range (two standard deviations above and below the average difference between trials) of

4.8 cm ( $-0.1 \pm 2.4$  cm) to 4.4 cm ( $0.2 \pm 2.2$  cm) and countermovement jumps from 8.4 cm ( $0.2 \pm 4.2$  cm) to 5.6 cm ( $0.0 \pm 2.8$  cm).

To compare stretch shortening cycle ability in vertical jumps, one needs to normalise for jumping ability, therefore simply comparing the difference between jumps would not be sufficient to draw accurate comparison between individuals of different ability. There are only two ways of normalising for jumping ability with countermovement- and squat jumps; expressing the difference between the two jumps either as (i) a ratio of change, or (ii) as a percentage change (Chapter 5). Comparative analysis has shown a linear relationship between the two ways of expression in all cases, hence it was decided to express this relationship as a percentage change, as this index has previously been related to maximal musculotendinous stiffness (101;151;152). The percentage jump height potentiation using the modified regional breakdown also improved from  $R = -0.08$  (95% CI:  $-0.32$ - $0.39$ ) and  $CV = 57.8$  (95% CI:  $43.7$ - $85.5$ ) to  $R = 0.76$  (95% CI:  $0.44$ - $0.93$ ) and  $CV = 32.4$  (95% CI:  $24.1$ - $49.3$ ). The limits of agreement also improved from a range of 55.2% ( $1.7 \pm 27.6\%$ ) to 23.4% ( $0.7 \pm 11.7\%$ ). Even though the variability was still reasonably high, this was a considerable improvement in the reliability of this measure with the revised analysis. On the basis of these findings the assumption was made that the other repeatable measures from Chapter 5 would also become more repeatable using this modified breakdown and therefore the modified regional breakdown would be used in subsequent studies.

The primary aim of this study was therefore to quantify the *in vivo* elastic properties of muscles and tendons in the lower body and to investigate their relationship with vertical jump performance and the potentiation of the stretch shortening cycle. A secondary aim of this study was to investigate and attempt to explain the underlying relationships between the various measures of stretch shortening cycle ability in the lower body.



## METHODS

### SUBJECTS

26 Male subjects with various training backgrounds, but with at least one year of resistance training experience were recruited for this study. Only 17 of these subjects completed the study. The subjects who were excluded from the study included 6 subjects who were unable to support the load with sufficient stability during the oscillation technique, 1 subject who could not tolerate more than 3 of the minimum 5 required loads on the oscillation test, and 2 subjects who aggravated old injuries and retired from the study for safety reasons. The remainder of the subjects ( $n = 17$ ) ranged between 20 and 32 years of age.

All subjects were informed about the nature and inherent risks involved in the study. The Ethics Committee of the Faculty of Health Sciences, University of Cape Town approved the study. Each subject completed a questionnaire regarding their medical, training and general information and further signed an informed consent form complying with the guidelines set by the American College of Sports Medicine (ACSM) (3), before participating in this study. Their general characteristics are summarized in Table 1. Body fat was expressed as a percentage (43) and sum of 7 skinfolds (7).

**Table 1:** General characteristics of the subject group ( $n = 17$ ).

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)
26.0 $\pm$ 3.6	174.5 $\pm$ 6.8	75.3 $\pm$ 8.5	75.5 $\pm$ 26.9	15.1 $\pm$ 4.0	63.9 $\pm$ 7.4

### EXPERIMENTAL DESIGN

The trial consisted of testing on three days, each separated by at least 48 hours. On each day, the testing was preceded by a standardized warm-up consisting of 5 - 10 minutes of continuous, self-paced, submaximal cycling on a stationary bike, specific stretches for the lower body and 2 sets of 10 repetition full squats with 20 kg (zero load on an Olympic bar).

The first day was used as a familiarization trial to teach the various techniques to the subjects and familiarize them with the procedures involved in the study. Particular attention was given to the familiarization of the oscillation test (151). Earlier studies have shown that certain individuals have difficulty in acquiring this technique and a significant amount of time was spent in familiarizing them with this test. Subjects were first familiarized with the modified oscillation test procedure (Chapter 4), as this was the test that would determine whether they were allowed to continue with the remainder of the study. They were given 4 - 6 trials per load to familiarize them with the oscillation test. The loads were increased incrementally to just below their expected maximum tolerable load.

If the subject could manage the oscillation technique, he was then familiarized with the remainder of the testing procedures i.e. the MVC and vertical jump tests. Subjects were given 3 - 6 trials per test to familiarize themselves with the techniques. After the familiarization day, the subjects had to visit the laboratory on two more occasions.

On the second day, body composition and the modified oscillation test (Chapter 4) were assessed. On the third testing day, the subject performed the MVC leg press and vertical jumps tests (Chapter 5).

To control for factors, which might affect muscle function, subjects were asked to refrain from any heavy or intense training during the testing period and also to avoid any lower body exercise at least 24 hours before each testing day. Subjects had at least 48 hours between the familiarisation trial and both testing days.

#### **THE MODIFIED OSCILLATION TEST FOR THE LOWER BODY**

The same procedure was used as previously described in Chapter 4 (p 169 - 170). However, two small modifications were made to the technique. The fact that the data of certain subjects in Chapter 4 had to be excluded, because they had too few data points, led to a modification to the protocol where the mass increments changed from 30 kg to 20 kg increments. This modification accommodated the weaker subjects and increased the scope of potential subjects. The starting mass was also altered from 70 kg to 80 kg. The test continued until the subjects were unable to support the load sufficiently for

accurate measuring of oscillations. The same exclusion criteria and controls were used as described in Chapter 4.

#### **TESTING FOR THE MVC ISOMETRIC LEG PRESS**

The same set-up and procedure for MVC isometric leg press was used as described in Chapter 5 (p 179 - 180).

#### **THE SSC TESTS FOR THE LOWER BODY**

The same techniques and controls for the testing of vertical jumps i.e. squat jumps and countermovement jumps, as discussed in Chapters 2 (p 124 - 126) and 5 (p 180), were used. All the force data, as for the MVC leg press, were sampled at a frequency of 2000 Hz using LabVIEW 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were used for further analysis.

For the analysis of force and EMG, the data were segmented using the modified force-time regional breakdown mentioned earlier in this chapter.

For the force- and EMG measures of squat jumps (Figure 1), the data between initiation of contraction (1) and take-off (2) were used. The flight time, which was calculated from the time difference between take-off (2) and landing (3) was used for the calculation of squat jump height.

The data between start of contraction (2) and take-off (3) were used for the force- and EMG measures of countermovement jumps (Figure 4). For the calculation of countermovement jump height, the flight time, which was calculated from the time difference between take-off (3) and landing (4), was used.

All force and EMG data were analysed using this regional breakdown. The same formulae used in Chapter 5 (p 182) were again used in the calculation of jump height for both tests.

In an attempt to have an alternative analysis of stretch shortening cycle potentiation, instantaneous power throughout both testing procedures i.e. squat jumps and

countermovement jumps, using the abovementioned regional breakdown and the formulae of Harman et al.(69) were calculated.

The vertical ground reaction force, which accelerates the body upwards in the jumps, was measured on the force plate throughout each test. As this force propels the entire body upwards, it can be assumed that the ground reaction force also acts through the body's centre of gravity, even though it is measured at the subject's feet (69). For the calculation of instantaneous power the force generated was multiplied by the velocity of movement. Hence, power was calculated using the following formula:

$$\text{Power (W)} = \text{Vertical ground reaction force (N)} \times \text{vertical velocity (m.s}^{-1}\text{)} \text{ (5;69) (a)}$$

Using the modified protocol for analysing vertical jump data, the force data were normalised for the bar-subject weight and force data were therefore regarded as true, as it is the net vertical force that effectively changes the vertical velocity and accelerates the body upwards (69).

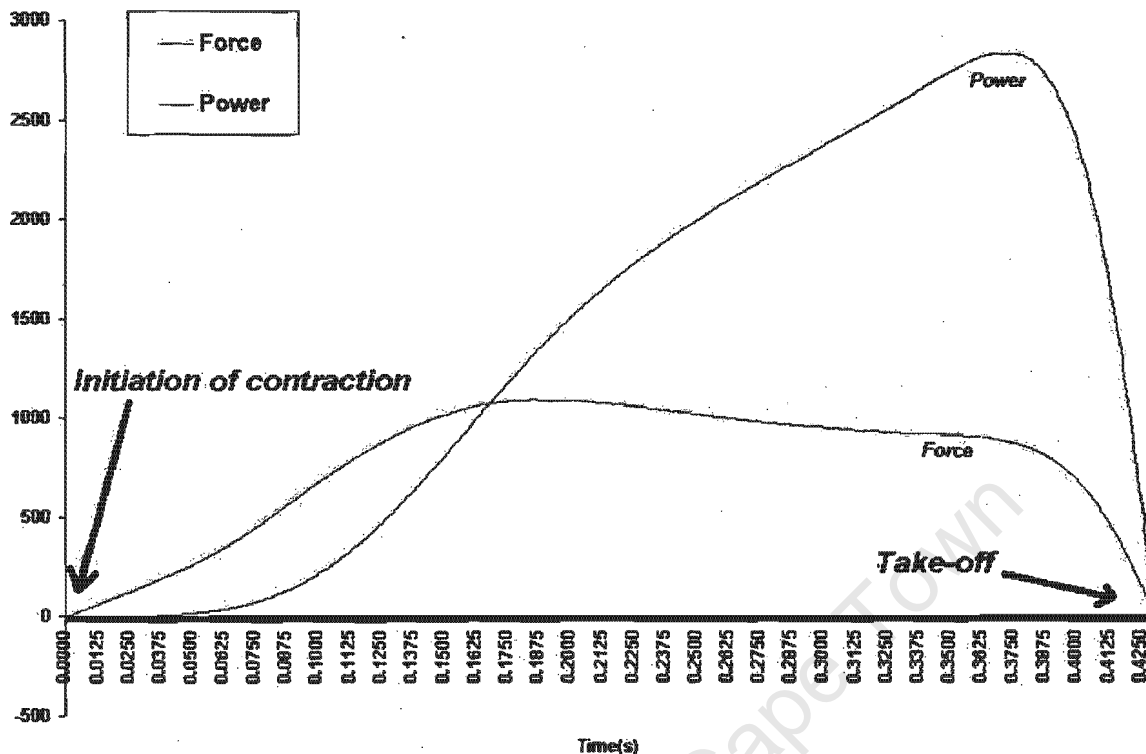
Vertical velocity had to be calculated using inverse principles and the premise that impulse equals a change in momentum. The following formula was used to calculate vertical velocity as the combined subject-bar mass remained constant:

$$\Delta V_v \text{ (m.s}^{-1}\text{)} = \{\text{Vertical ground reaction force (N)} \times \text{time period (s)}\} / \text{bar-subject mass (kg)} \text{ (69) (b)}$$

Where  $\Delta V_v$  = the change in vertical velocity.

Starting with zero at the beginning of the jump, the absolute vertical velocity was continuously updated by adding the change in vertical velocity ( $\Delta V_v$ ) of each 0.0005 s epoch to the vertical velocity of the preceding interval (69).

Instantaneous power was calculated throughout as the product of vertical ground reaction force and absolute vertical velocity (69); the peak value was defined as the maximal explosive muscle power (159). Peak power and average power were calculated within the regional breakdown mentioned earlier (Figures 1, 4 and 5).



**Figure 5:** The calculation of instantaneous power is calculated throughout the jump, using the modified regional breakdown. (\* Graph from original data)

## ELECTROMYOGRAPHY (EMG)

For the modified oscillation test, surface EMG was measured to identify and exclude trials, which were confounded by muscle recruitment (149;151;155;156). The same procedure was used as described in Chapter 4 (p 170).

For the vertical jumps the same sampling strategy, preparation, muscle group and procedure was used for EMG analysis, as described previously in Chapter 5 (p 182 - 184). The only modifications made in this protocol were the regional breakdown mentioned earlier (Figures 1 and 4).

It must be noted that both legs contributed to the force generated in the trials whereas the EMG related measures were only measured on the right leg. However, as vertical jumping is assumed to be a bilaterally symmetrical movement (63), the right limb was selected as

being representative of both sides (78) and because countermovement jumps were compared to squat jumps in the same individuals, this was accepted as a systematic error.

### STATISTICAL ANALYSIS

Statistical analysis of the data was performed using Statistica Version 6.0 statistical analysis- (Statsoft Inc., Tulsa, U.S.A.) and Graphpad Prizm Version 3.0 software (Graphpad software Inc., San Diego, California, U.S.A.). For relationships between variables the Pearson's product moment correlation coefficient was used. A correlation matrix was calculated using all the test data. A T-test for dependent variables was used to establish significant differences between variables. Statistical significance was accepted at  $p < 0.05$ .

### RESULTS

#### MUSCULOTENDINOUS STIFFNESS VS VERTICAL JUMP DATA

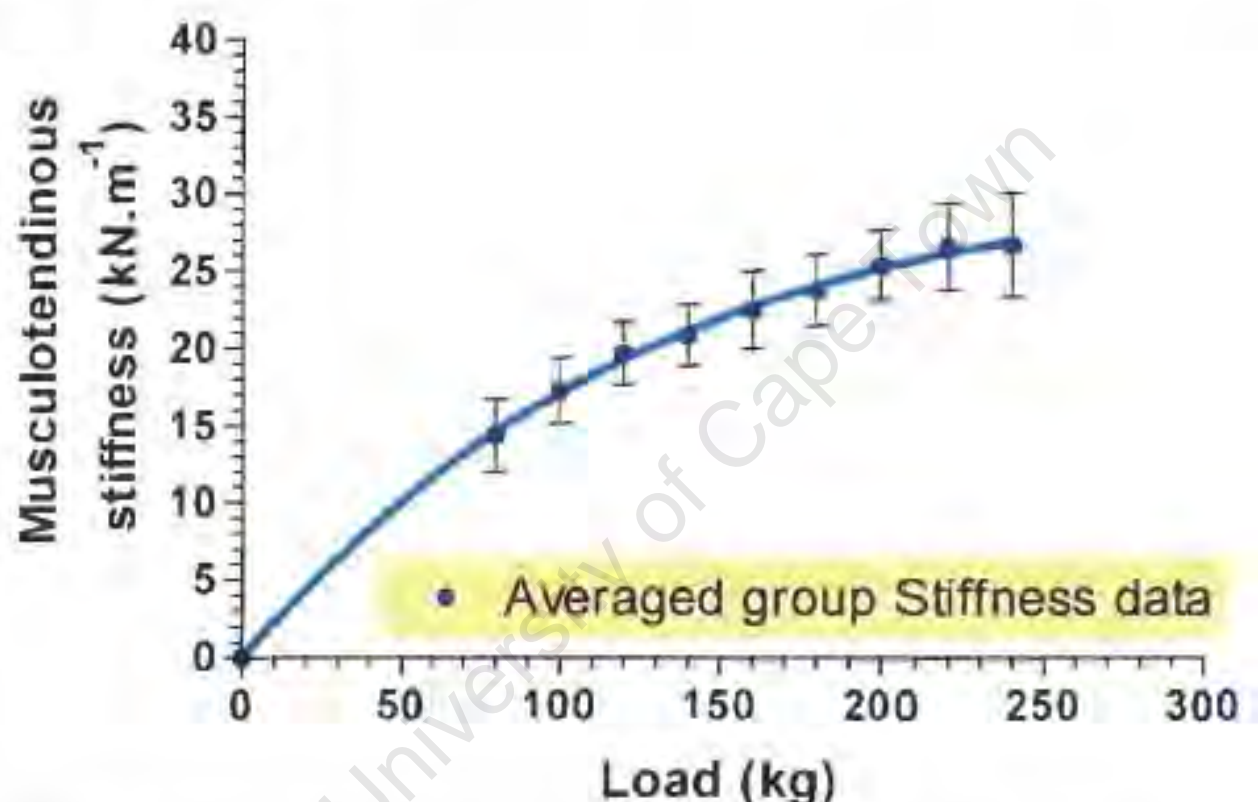
The range of musculotendinous stiffness values, maximal (tendon) and submaximal, of the subjects tested ( $n = 17$ ) in this study using the modified oscillation technique are shown in Table 2.

**Table 2:** *Musculotendinous stiffness values ( $\text{kN.m}^{-1}$ ) of the subjects over the range of absolute loads and the maximal predicted stiffness of the subjects ( $n = 17$ ).*

Subject	80kg	100kg	120kg	140kg	160kg	180kg	200kg	220kg	240kg	260kg	280kg	300kg	Maximal stiffness
Down1	11.8	13.7	16.9	18.7	19.5	22.0	23.2	24.6	-	-	-	-	33.7
Down2	10.8	13.4	18.1	18.7	20.9	23.0	-	-	-	-	-	-	25.8
Down3	15.5	18.5	21.6	23.7	23.7	27.1	27.9	29.3	-	-	-	-	35.6
Down4	10.9	14.2	17.0	19.1	19.8	19.9	22.5	23.3	-	-	-	-	24.3
Down5	13.0	16.5	19.1	19.9	25.8	26.4	27.6	28.0	-	-	-	-	31.8
Down6	12.6	17.3	17.4	19.4	22.7	-	-	-	-	-	-	-	37.0
Down7	12.8	16.8	18.9	21.6	23.9	22.4	-	-	-	-	-	-	24.0
Down8	11.7	15.8	17.6	17.8	18.1	22.3	23.8	23.3	24.4	27.0	29.0	30.0	39.6
Down9	16.1	18.0	20.7	22.1	23.1	24.0	-	-	-	-	-	-	26.4
Down10	13.7	16.2	20.9	23.1	25.1	26.2	-	-	-	-	-	-	28.9
Down11	19.4	19.4	21.7	23.2	24.3	23.0	27.5	27.4	-	-	-	-	28.3
Down12	16.5	20.3	21.5	20.6	21.9	23.9	-	-	-	-	-	-	23.0
Down13	16.7	19.1	22.2	22.8	25.3	28.2	-	-	-	-	-	-	38.1
Down14	15.8	18.4	20.9	23.1	25.0	24.2	-	-	-	-	-	-	26.2
Down15	14.9	17.8	17.2	18.5	20.8	-	-	-	-	-	-	-	22.3
Down16	16.7	20.2	21.4	20.9	18.6	21.2	-	-	-	-	-	-	20.5
Down17	15.2	18.8	21.6	22.8	24.5	22.9	25.4	30.0	29.0	-	-	-	33.5
Average	14.4	17.3	19.7	20.9	22.5	23.8	25.4	26.6	26.7	27.0	29.0	30.0	29.4
± SD	± 2.4	± 2.1	± 2.0	± 2.0	± 2.5	± 2.3	± 2.3	± 2.8	± 3.3				± 6.0

The tendon stiffness values ranged from 20.5 - 39.6 kN.m<sup>-1</sup> averaging  $29.4 \pm 6.0$  kN.m<sup>-1</sup>. The subjects managed on average between 6 - 8 loads before the data became unstable.

Figure 6 shows that the group data once again followed the classic curvilinear model, which was found in the earlier studies (Chapters 3 and 5) and has also been shown in previous research (132;151;153;156). The results of this study also confirmed that, as with the earlier studies (Chapters 3 and 5), the individual subject data became more variable at the heavier loads.



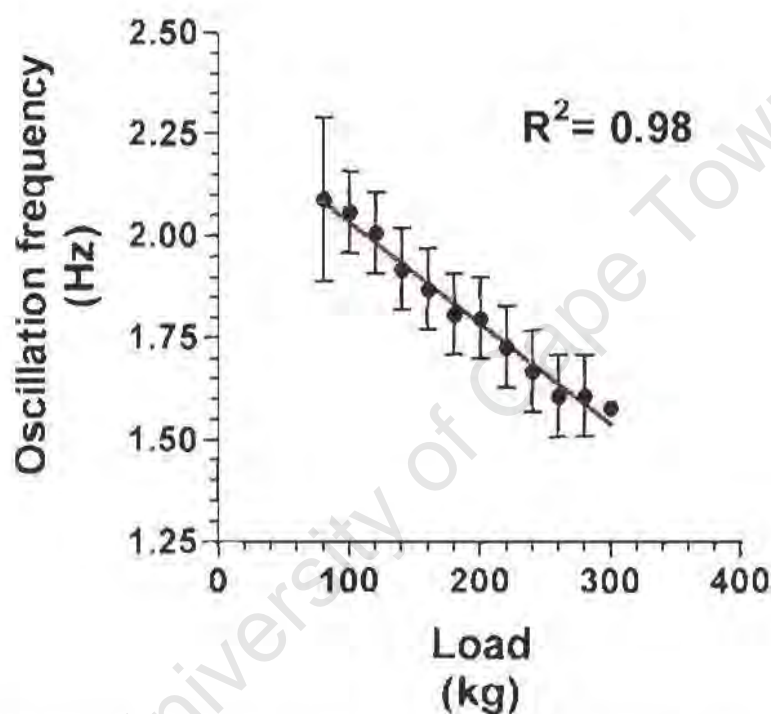
**Figure 6:** The grouped average stiffness-load relationship data of the subjects used in this study ( $n = 17$ ). Only the data up to 240kg were plotted as only 1 subject managed the heavier loads.

The damped natural frequency of oscillation (Figure 7 and Table 3) also decreased as the load on the lower body was increased. This has also frequently been shown in the earlier studies (Chapters 3 and 5). The oscillation frequency data ranged from 1.58 - 2.09 Hz.



**Table 3:** The average damped natural frequency data at the relative loads using the NAMS unit  
(n x 6 trials per load)

Absolute load (kg)	Damped Natural Frequency (Hz)
80	2.09 ± 0.19
100	2.06 ± 0.14
120	2.01 ± 0.12
140	1.92 ± 0.10
160	1.87 ± 0.12
180	1.81 ± 0.10
200	1.80 ± 0.10
220	1.73 ± 0.10
240	1.67 ± 0.10
260	1.61 ± 0.05
280	1.61 ± 0.09
300	1.58 ± 0.04



**Figure 7:** The response of the damped natural frequency of oscillation of the muscle-tendon complex under incremental loading conditions (n = 17).

Table 4 shows the range of vertical jump test data that was used for determining the stretch shortening cycle potentiation of vertical jump performance.

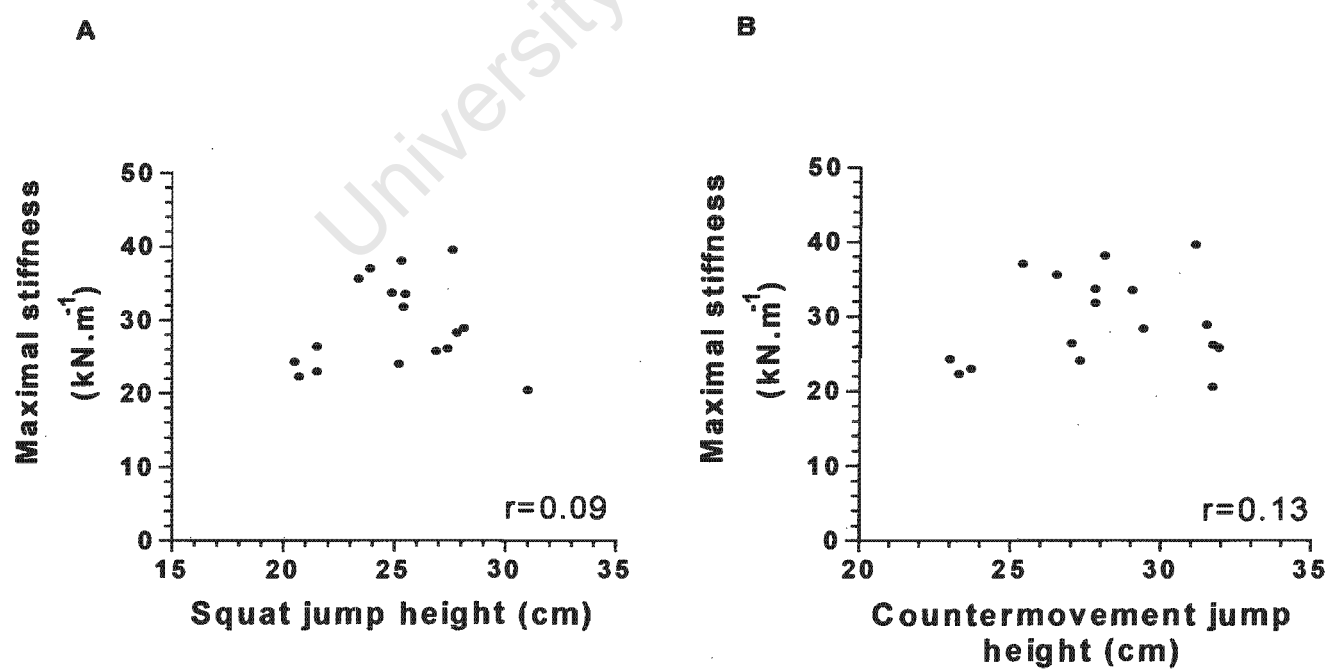


**Table 4:** Data collected during the vertical jump tests used in calculation of stretch shortening cycle performance potentiation (*n* = 17).

Subject	SJ (cm)	CMJ (cm)	SJ Impulse (Ns)	CMJ Impulse (Ns)	SJ IEMG (%MVC)	CMJ IEMG (%MVC)	SJ F: EMG (N/%)	CMJ F: EMG (N/%)	SJ AP (W)	CMJ AP (W)	SJ PP (W)	CMJ PP (W)
Down1	24.9	27.8	216.8	350.0	11.0	16.7	19.7	20.9	657	1165	1694	2744
Down2	26.9	31.9	276.9	434.4	16.2	23.4	17.1	18.5	917	1776	2741	3716
Down3	23.4	26.5	211.0	337.1	15.9	20.3	13.3	16.6	705	1305	1702	2712
Down4	20.5	23.0	192.2	301.1	14.7	16.1	13.1	18.7	424	1077	1317	2384
Down5	25.4	27.8	222.0	363.5	17.0	18.4	13.1	19.7	605	1374	1645	2808
Down6	23.9	25.4	187.2	265.4	12.9	15.8	14.5	16.8	440	918	1486	2126
Down7	25.2	27.3	208.6	317.7	12.9	12.6	16.2	25.2	589	1123	1392	2532
Down8	27.6	31.1	244.7	416.1	7.4	10.3	33.2	40.4	857	1836	1969	3285
Down9	21.5	27.0	183.0	315.0	15.5	13.2	11.8	23.9	467	1222	1301	2589
Down10	28.1	31.5	240.4	398.5	7.2	8.5	33.2	47.1	840	1760	1900	3207
Down11	27.8	29.4	211.1	343.2	12.1	15.8	17.4	21.8	887	1743	1985	3188
Down12	21.5	23.7	196.6	314.2	9.9	15.1	19.8	20.8	655	1155	1376	1907
Down13	25.3	28.1	223.0	349.3	12.2	15.6	18.2	22.3	828	1238	2029	3039
Down14	27.4	31.7	233.2	393.5	16.4	19.9	14.2	19.8	614	1640	2136	3601
Down15	20.7	23.3	170.8	276.4	13.7	18.0	12.5	15.3	556	986	1129	1865
Down16	31.0	31.7	230.5	327.5	9.9	16.4	23.2	20.0	1124	1393	2437	2825
Down17	25.5	29.0	223.0	385.6	14.0	21.0	16.0	18.4	714	1713	1640	2914
Average	25.1	28.0	215.9	346.4	12.9	16.3	18.0	22.7	699	1378	1758	2791
± SD	± 2.9	± 3.0	± 26.0	± 47.6	± 3.0	± 3.8	± 6.5	± 8.4	± 189	± 308	± 429	± 532

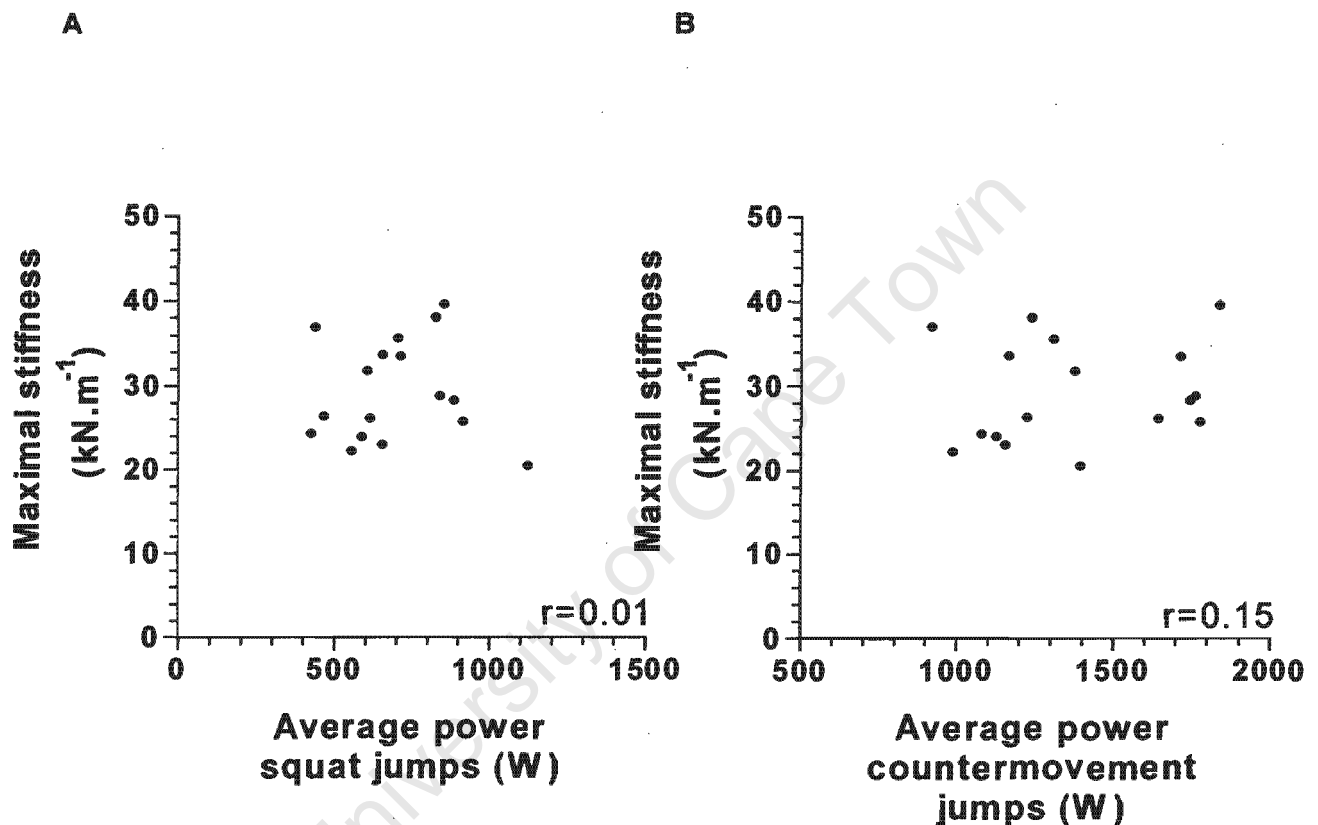
**Index:** SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG, F: EMG = Force to EMG ratio, AP = average power, PP = peak power

Figure 8 shows the scatterplot between maximal (tendon) stiffness and vertical jump heights in (A) squat jumps and (B) countermovement jumps respectively. It is clear that there is no relationship between tendon stiffness and either of these two variables (SJ – *r* = 0.09 and CMJ – *r* = 0.13).



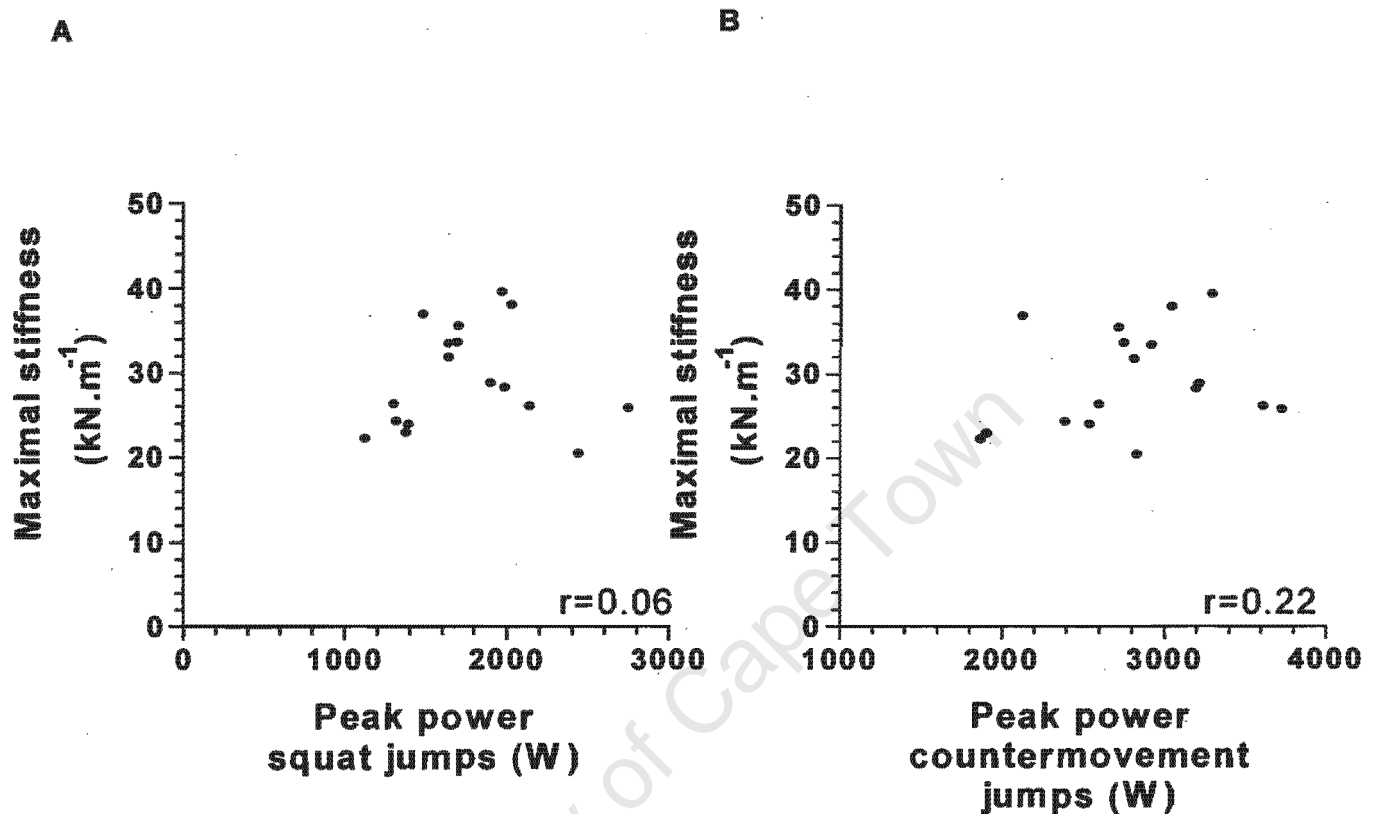
**Figure 8:** A scatterplot of maximal (tendon) stiffness and vertical jump height in squat jumps (A) and countermovement jumps (B) (*n* = 17).

Figure 9 shows the scatterplot between maximal (tendon) stiffness and average power in squat jumps (A) and countermovement jumps (B) respectively. It is clear, as with the jump heights, that there is no relationship between tendon stiffness and either of these two variables (SJ –  $r = 0.01$  and CMJ –  $r = 0.15$ ).



**Figure 9:** A scatterplot of maximal (tendon) stiffness and average power generated in squat jumps (A) and countermovement jumps (B) ( $n = 17$ ).

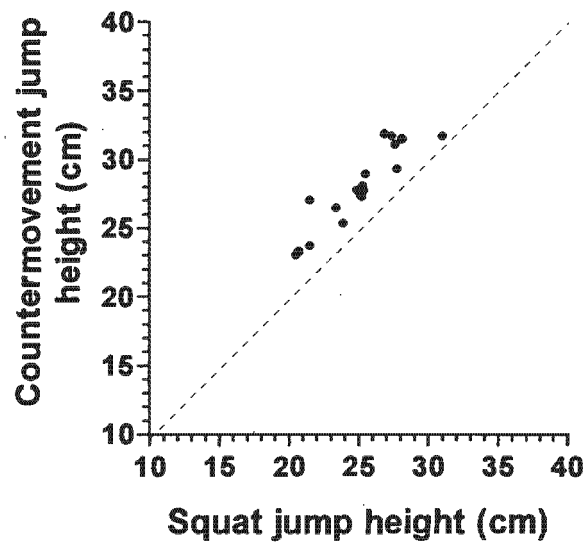
Figure 10 shows the scatterplot between maximal (tendon) stiffness and peak power in squat jumps (A) and countermovement jumps (B) respectively. It is clear that there is also no relationship between tendon stiffness and either of these two variables (SJ –  $r = 0.06$  and CMJ –  $r = 0.22$ ).



**Figure 10:** The relationship between maximal (tendon) stiffness and peak power generated in squat jumps (A) and countermovement jumps (B) ( $n = 17$ ).

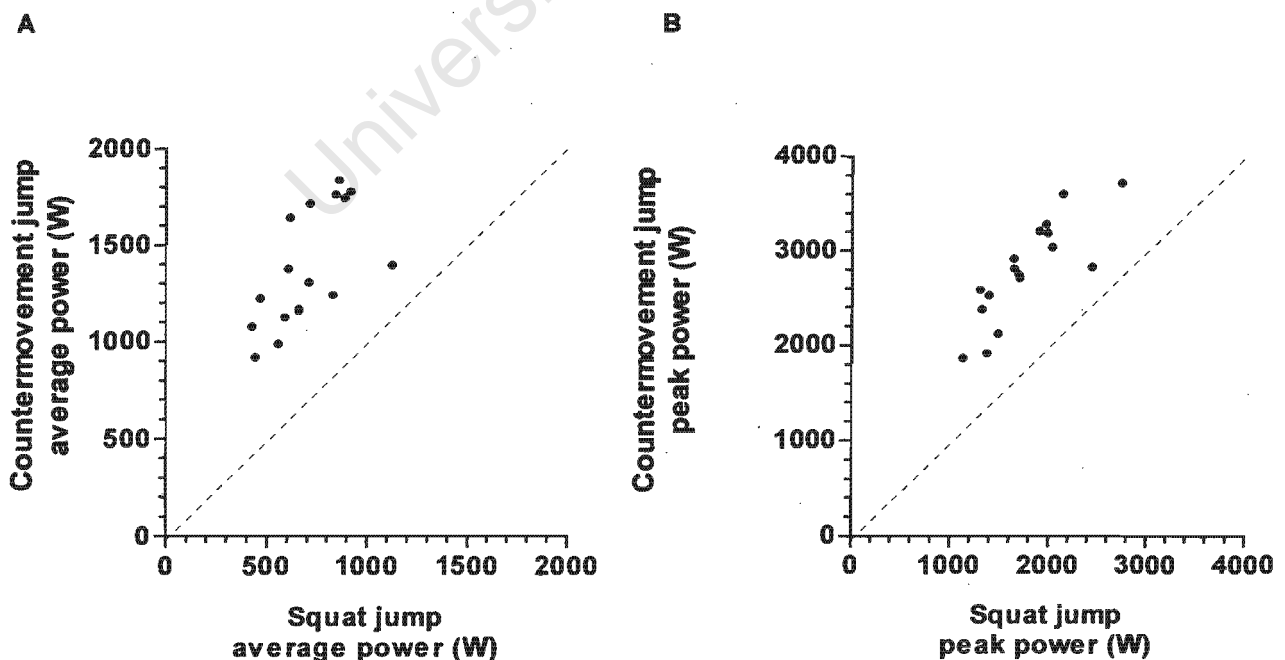
### **STRETCH SHORTENING CYCLE POTENTIATION OF VERTICAL JUMP DATA**

When comparing countermovement and squat jump height data, it was found that countermovement jumps were significantly higher ( $p < 0.0000001$ ) than squat jumps. This is further visible in Figure 11, indicating that significant jump height potentiation occurred utilising the stretch shortening cycle of muscle function compared to isolated concentric jumps (squat jumps).



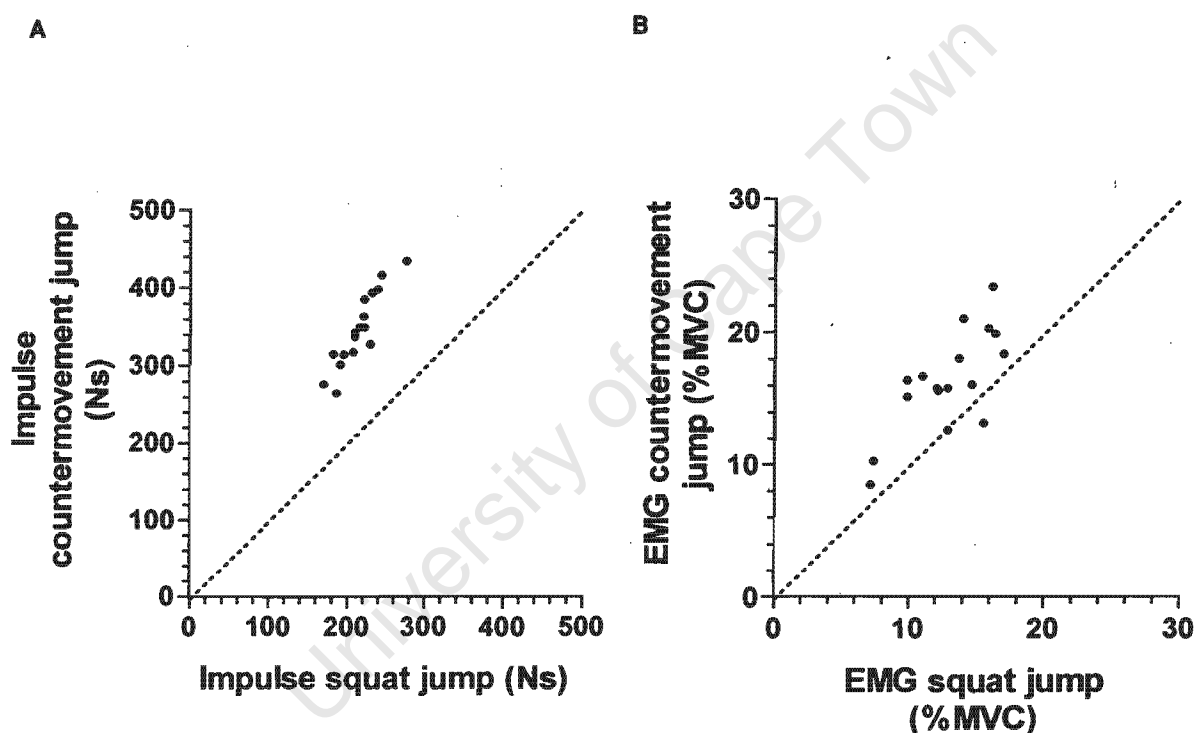
**Figure 11:** The comparison between the vertical jump heights in squat jumps and counter movement jumps. The data points are consistently above the line of equality (dotted line) indicating stretch shortening cycle potentiation ( $n = 17$ ).

When comparing counter movement and squat jump average and peak power data, it was also found that both average- ( $p < 0.0000001$ ) and peak power ( $p < 0.0000001$ ) were significantly higher in counter movement jumps than squat jumps. This is further visible in Figure 12, indicating that significant potentiation in both average power (A) and peak power (B) occurred utilising the stretch shortening cycle of muscle function compared to isolated concentric jumps (squat jumps).



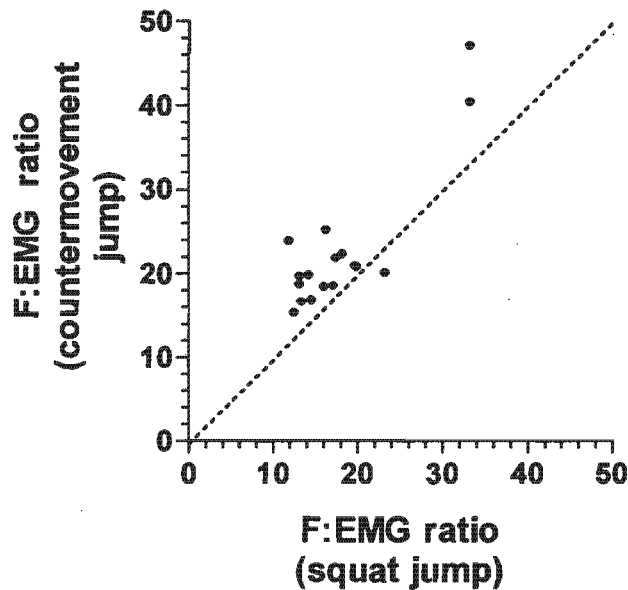
**Figure 12:** The comparison between average power (A) and peak power (B) generated in squat jumps and counter movement jumps. The data points are consistently above the line of equality (dotted line) indicating stretch shortening cycle potentiation ( $n = 17$ ).

When comparing countermovement and squat jump impulse (defined by area under the force-time curve) (10;11) and EMG (integrated EMG data), it was also found that both impulse- ( $p < 0.0000001$ ) and EMG ( $p < 0.00005$ ) were significantly higher in countermovement jumps than squat jumps. This is also further visible in Figure 13, indicating that significant potentiation in both impulse (A) and EMG (B) occurred utilising the stretch shortening cycle of muscle function compared to isolated concentric jumps (squat jumps). Average force, which relates to impulse, followed the same pattern and was significantly higher ( $p < 0.00002$ ) in countermovement jumps ( $754.8 \pm 123.7$  N) than squat jumps ( $587.1 \pm 131.8$  N).



**Figure 13:** The comparison between impulse (A) and EMG (B) in squat jumps and countermovement jumps. The data points are consistently above the line of equality (dotted line) indicating stretch shortening cycle potentiation ( $n = 17$ ). (\* 2 Subjects in (B) did not show potentiation using these data)

When comparing countermovement and squat jump F: EMG data, it was also found that the F: EMG ratio was significantly higher in countermovement jumps ( $p < 0.0003$ ) than squat jumps. This is visible in Figure 14, indicating that significant potentiation in the F: EMG ratio occurred utilising the stretch shortening cycle of muscle function compared to isolated concentric jumps (squat jumps).



**Figure 14:** The comparison between F: EMG ratios in squat jumps and countermovement jumps. The data points are consistently above the line of equality (dotted line) indicating stretch shortening cycle potentiation ( $n = 17$ ). (\* 1 Subject did not show potentiation)

Table 5 shows the stretch shortening cycle potentiation measures, expressed as a percentage (%), determined from the vertical jump data (Table 4).

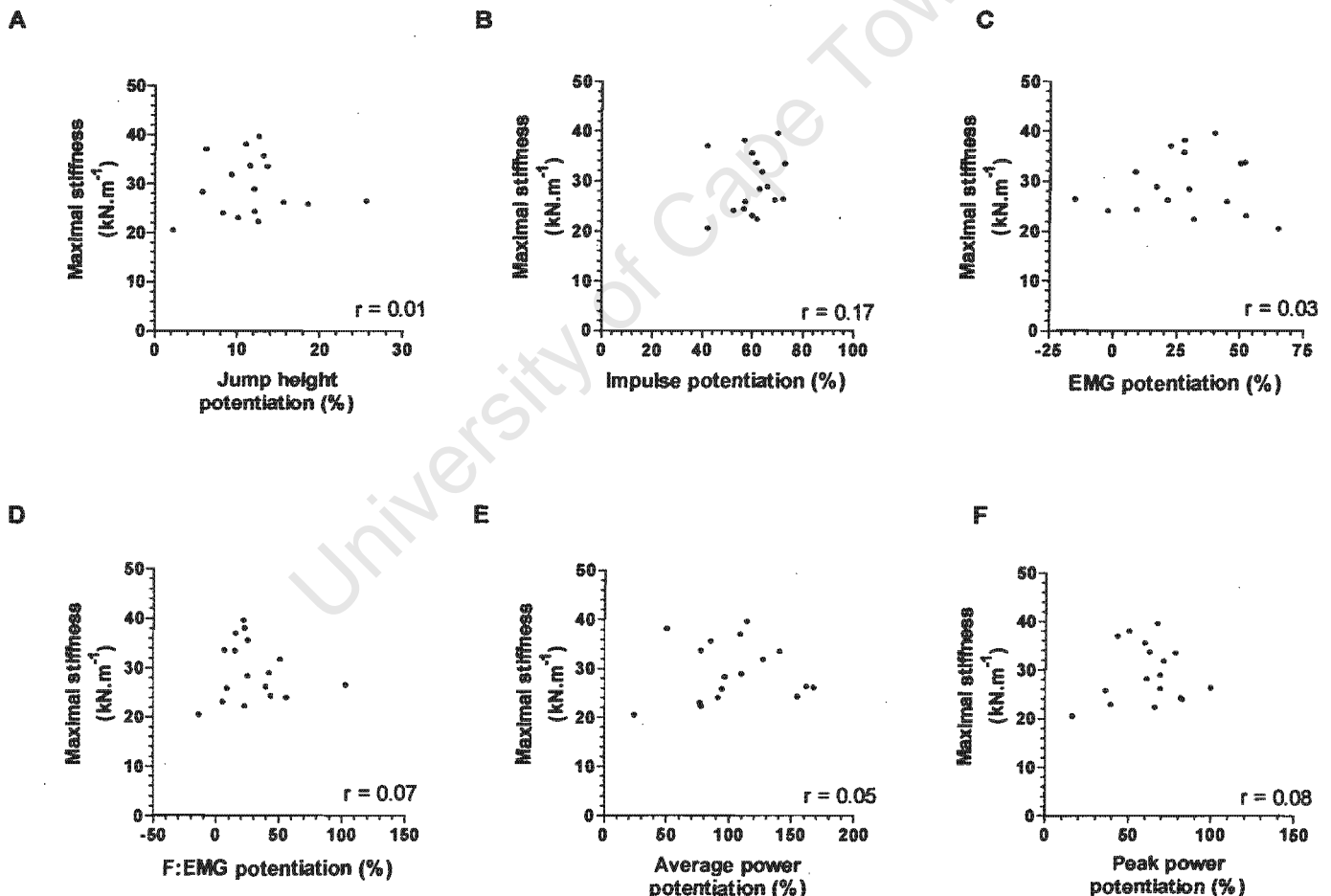
**Table 5:** Stretch shortening cycle potentiation measures used in vertical jump testing ( $n = 17$ ).

Subject	CMJ-SJ height Potentiation (%)	CMJ-SJ Impulse Potentiation (%)	CMJ-SJ %IEMG Potentiation (%)	CMJ-SJ F: EMG Potentiation (%)	CMJ-SJ AP Potentiation (%)	CMJ-SJ PP Potentiation (%)
Down1	11.6	61.4	51.7	6.4	77.3	62.0
Down2	18.6	56.9	44.8	8.4	93.7	35.6
Down3	13.2	59.8	27.5	25.3	85.1	59.3
Down4	12.2	56.7	9.2	43.4	154.0	81.0
Down5	9.4	63.7	8.6	50.8	127.1	70.7
Down6	6.3	41.8	22.6	15.6	108.6	43.1
Down7	8.3	52.3	-2.2	55.7	90.7	81.9
Down8	12.7	70.0	39.8	21.6	114.2	66.8
Down9	25.6	72.1	15.1	102.8	161.7	99.0
Down10	12.1	65.8	17.0	41.7	109.5	68.8
Down11	5.8	62.6	29.7	25.3	96.5	60.6
Down12	10.2	59.8	52.2	5.0	76.3	38.6
Down13	11.1	56.6	27.8	22.6	49.5	49.8
Down14	15.7	68.7	21.2	39.2	167.1	68.6
Down15	12.6	61.8	31.7	22.8	77.3	65.2
Down16	2.3	42.1	65.0	-13.9	23.9	15.9
Down17	13.7	72.9	49.9	15.3	139.9	77.7
<b>Average</b>	<b>11.8</b>	<b>60.3</b>	<b>28.3</b>	<b>28.7</b>	<b>103.1</b>	<b>61.4</b>
<b>± SD</b>	<b>± 5.2</b>	<b>± 9.0</b>	<b>± 21.1</b>	<b>± 26.2</b>	<b>± 38.8</b>	<b>± 20.1</b>

**Index:** SJ = squat jumps, CMJ = countermovement jumps, IEMG = integrated EMG,  
F: EMG = Force to EMG ratio, AP = average power, PP = peak power

## **MUSCULOTENDINOUS STIFFNESS VS STRETCH SHORTENING CYCLE POTENTIATION**

Figure 15 shows the scatterplots between maximal (tendon) stiffness and the various measures used in calculating stretch shortening cycle potentiation (Table 5) using vertical jumps. It is clear that there also is no relationship between tendon stiffness and any of these variables. For example no relationship was found between maximal tendon stiffness and (A) jump height potentiation ( $r = 0.01$ ), (B) impulse potentiation ( $r = 0.17$ ), (C) EMG potentiation ( $r = 0.03$ ), (D) F: EMG ratio potentiation ( $r = 0.07$ ), (E) average power potentiation ( $r = 0.05$ ), and (F) peak power potentiation ( $r = 0.08$ ). All the submaximal stiffness values were also compared to the entire vertical jump data. Once again there were no relationships between submaximal stiffness and any of these variables.

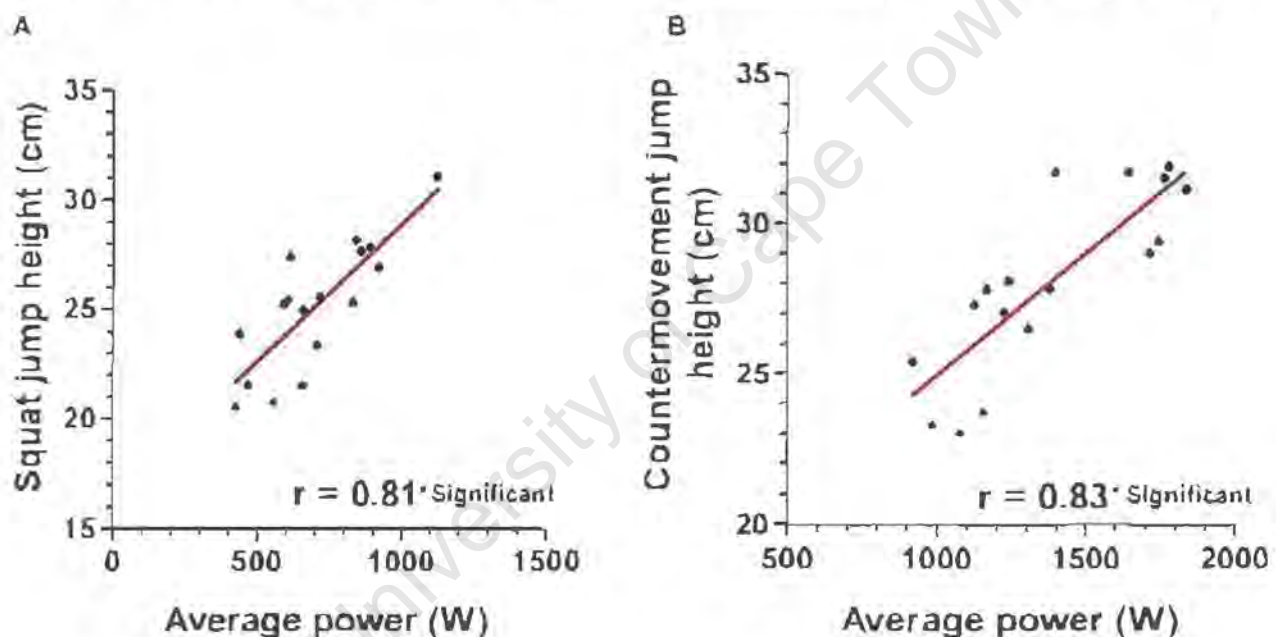


**Figure 15:** The scatterplots of maximal (tendon) stiffness and the various measures of potentiation in vertical jumps. A – Jump height potentiation, B – Impulse potentiation, C – EMG potentiation, D – F: EMG potentiation, E – Average power potentiation, F – Peak power potentiation ( $n = 17$ ).

## VERTICAL JUMP HEIGHT AND PERFORMANCE DATA

The average- (Figure 16) and peak power output data (Figure 17) were compared using the modified regional breakdown and the formulae of Harman et al. (69), with the calculated vertical jump height in both squat jumps and countermovement jumps on the NAMS unit.

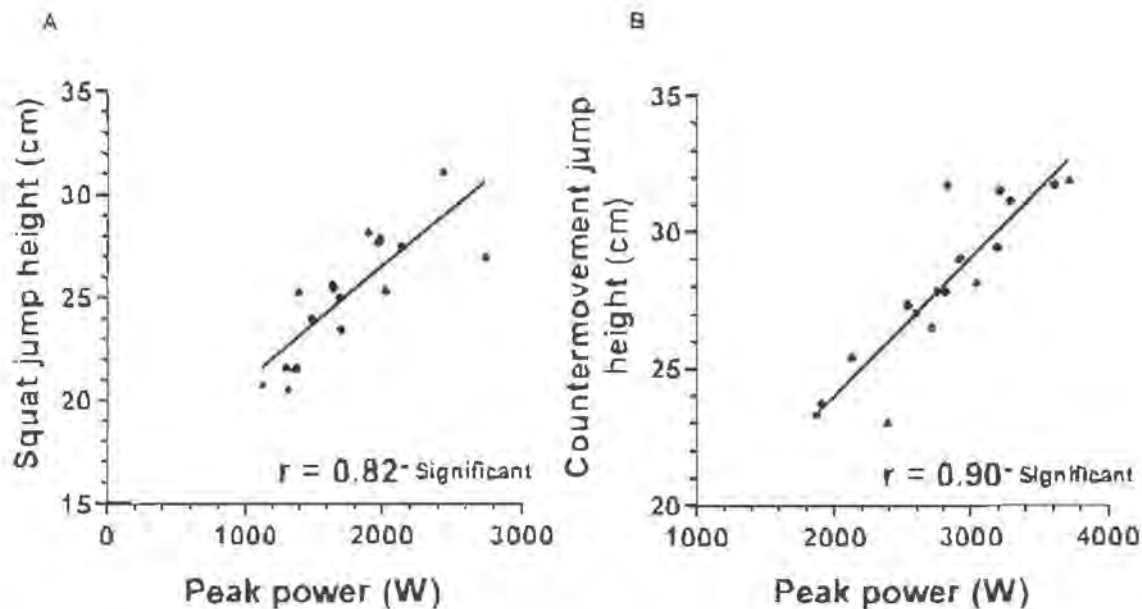
Figure 16 shows the relationship between squat jumps (A) and countermovement jumps (B) and their respective average power data. Both squat jump height ( $r = 0.81$ ,  $p < 0.0001$ ) and countermovement jump height ( $r = 0.83$ ,  $p < 0.0001$ ) were highly correlated to the average power output of their respective force-time data (Figure 16).



**Figure 16:** The relationship between vertical jump height and average power in squat jumps (A) and countermovement jumps (B) ( $n = 17$ ).

Figure 17 shows the relationship between squat jumps (A) and countermovement jumps (B) and their respective peak power data. Both (A) squat jump height ( $r = 0.82$ ,  $p < 0.0001$ ) and (B) countermovement jump height ( $r = 0.90$ ,  $p < 0.0001$ ) were highly correlated to the peak power output of their respective force-time data (Figure 17).





**Figure 17:** The relationship between vertical jump height and peak power in squat jumps (A) and countermovement jumps (B) ( $n = 17$ ).

Net impulse and average force data were also compared to the vertical jump height data. Net impulse of their respective force plate data correlated well with both squat jump height ( $r = 0.75$ ,  $p < 0.0005$ ) and countermovement jump height ( $r = 0.81$ ,  $p < 0.0001$ ). Similar relationships were encountered with the average force data; squat jumps ( $r = 0.73$ ,  $p < 0.0009$ ) and countermovement jumps ( $r = 0.77$ ,  $p < 0.0003$ ).

## DISCUSSION

### MUSCULOTENDINOUS STIFFNESS AND STRETCH SHORTENING CYCLE POTENTIATION

The primary aim of this study was to investigate the relationship between maximal musculotendinous or tendon stiffness, as determined by the modified oscillation technique, and stretch shortening cycle ability established by vertical jumps.

Numerous techniques have been used to determine musculotendinous stiffness. Different types of muscle contraction, intensity levels, joint positions, protocols and mathematical equations have been used to quantify the elastic or stiffness properties of the muscle-tendon complex (152). The diversity of muscle architecture additionally affects the

variation in musculotendinous stiffness registered amongst different muscle groups (128). This contributes to a wide variation in stiffness data from different studies. Care should therefore be taken when comparing absolute values in the literature derived from different protocols (152). In accordance with this the results from this study have been related to only those studies, which used a similar procedure of measurement.

The subjects in this study managed on average between 6 and 8 loads on the modified oscillation procedure and generated a wide range of musculotendinous stiffness values. The maximal or tendon stiffness data ranged from 20.5 - 39.6 kN.m<sup>-1</sup> averaging 29.4 ± 6.0 kN.m<sup>-1</sup>, which is consistent with the data gathered in the earlier studies as well as values observed in the literature using similar techniques of measurement (Chapter 3, p 146 - 147). The oscillation frequency data were also consistent with that found in the literature (Chapter 3, p 147) and are similar to those values generated in the earlier studies. The oscillation frequency data ranged between 1.58 - 2.09 Hz over the various loads.

The major finding in this study, however, was that there was no relationship between both maximal (tendon) and submaximal musculotendinous stiffness measures, and any measurements of potentiation associated with the stretch shortening cycle. A superficial analysis suggests that this finding contradicts the previous literature on this topic.

However, when the extent of the associations between variables found in the other studies is examined in more detail, an alternative argument can be developed. While there were significant relationships between tendon stiffness and stretch shortening cycle performance improvement (97;101;151;152), when reviewing their correlation coefficients, the majority of the studies (Chapter 1, Table 3, p 80) showed relatively low correlations and therefore low predictability of one variable by another. For example, Kubo et al. (97), Walshe et al. (151), Kubo et al. (101) and Watsford et al. (152) could only explain 30%, 25%, 21% and 42% of the variance in stretch shortening cycle performance enhancement respectively by differences in musculotendinous stiffness. Therefore, these studies could only explain between 21 and 42% of the variance in performance gain. Even though the relationships were significant in all the abovementioned studies, the associations are rather weak.

The data in this study, using multiple muscle groups, in a multi-joint testing procedure similar to that of Walshe et al. (151), question these findings and query the ability of the isolated measurement of tendon stiffness to explain stretch shortening cycle performance potentiation in the lower body.

During the test using the oscillation technique it is important that subjects maintain equilibrium in the muscle-tendon complex (i.e. the force output required to balance the load) before, during and after perturbation to prevent the antagonistic muscles from hindering the natural oscillations induced by the perturbation (32). The main consequence of maintaining a steady-state contraction against a load and not intervening in any way is that the mechanical characteristics of the muscle-tendon system would promote the return of static equilibrium or the return of the joint to its original position after the perturbation (106).

The skill required to maintain static equilibrium and control during the oscillation test was lacking in 23% of the subjects initially recruited and screened for this study. Even those subjects who had managed the technique seemed to have had difficulty maintaining the very light and very heavy loads. This observation is supported in the literature (106). However, the stringent control measures incorporated into the study design (Chapter 2, p 121 - 123, Chapter 4, p 169 - 170) ensured that the data were reliable and could be used to predict maximal- or tendon stiffness as accurately as possible within the limitations of the technique utilised.

## **MUSCULOTENDINOUS STIFFNESS AND VERTICAL JUMP PERFORMANCE**

It would seem there is conflicting evidence with regards to musculotendinous stiffness and its effect on vertical jump performance parameters (Chapter 1, Table 3, p 80). Although musculotendinous stiffness does not seem to be related to vertical jump height, in any of the jumps (101;151;152), it does seem to be associated with the rate of force development, particularly in the squat jumps or concentric movements (151;152). This is logical, because with a stiffer tendon the force generated in the muscles would be transferred more rapidly to the bones to create movement, and the transition between onset of contraction and movement would be accelerated (27) leading to a more powerful

contraction (155). Considering that the rate of force development was not calculated in this study, it was therefore assumed any relationship between maximal (tendon) stiffness and the rate of force development, would be expressed in its relation to the power output of the involved musculature. Therefore, it was expected that in the squat jump condition, there would be a relationship between stiffness and power output, which relates to the rate of force development.

The musculotendinous stiffness data was compared at all levels with vertical jump heights, average power outputs, peak power outputs, and all the other individual kinetic variables in both squat jump and countermovement jump conditions. None of these comparisons showed any relationship between maximal (tendon) or submaximal musculotendinous stiffness values of the lower body musculature. These findings dispute what has previously been shown in the literature and indicate that there is no *de facto* relationship between maximal (tendon) and submaximal musculotendinous stiffness, and any of the vertical jump- and stretch shortening cycle performance parameters in the lower body, at least in the way they were measured in this study.

## **MUSCLE FUNCTION DURING COUNTERMOVEMENT JUMPS AND SQUAT JUMPS**

### **JUMP HEIGHT**

This study showed a significant ( $p < 0.0000001$ )  $12 \pm 5\%$  increase in countermovement jump height ( $28.0 \pm 3.0$  cm) over squat jump height ( $25.1 \pm 2.9$  cm). This has also been shown in many other studies (Chapter 1, Table 1, p 15).

### **FORCE AND POWER**

Using the methodology described in Chapter 5, it was not possible to accurately differentiate between eccentric and concentric phases of contraction (Chapter 5). Therefore the muscle performance data described in this study are representative of the combined eccentric (lengthening) and concentric (shortening) phases of contraction. Komi

(91), however, states that during the stretch shortening cycle one cannot regard eccentric and concentric phases separately as the one significantly influences the other. Nonetheless, as previously discussed in the literature review, during stretch shortening cycle movements it is the eccentric contraction that is instrumental in the enhancement of the subsequent concentric contraction, and the propulsion phase of vertical jumps is mainly responsible for increasing jump height. One therefore has to assume that it is predominantly the effective modification of the concentric phase of the stretch shortening cycle muscle action that results in the functional improvements.

In this study average force was  $32 \pm 24\%$  higher in countermovement jumps than in squat jumps. Similar, yet more pronounced improvements were noted in average and peak power; average power increased by  $103 \pm 39\%$  and peak power improved by  $61 \pm 20\%$ . These findings seem to be consistent with the data reported in the literature and furthermore additionally verified, as seen in the literature (Chapter 1, Table 1, p 15) that power was more affected by the stretch shortening cycle muscle action than force output.

## EMG

By eccentrically stretching muscles as part of the stretch shortening cycle, muscle activation, represented by surface EMG, increases appreciably when compared to isometric and concentric contractions (144). During the eccentric phase of stretch shortening cycle activities, there is considerable activation of the involved musculature followed by significantly lower (91;137) or unchanged (138-140) EMG during the subsequent concentric phase. Stretch reflex contributions are in most cases brought about during the latter part of the eccentric phase of stretch shortening cycle activities (93). Faster eccentric stretching of the active muscles leads to more electrical activation and by implication more muscle recruitment, which increases muscle stiffness. This increase in muscle stiffness seems to be an outcome of the aforementioned reflex activity in conjunction with pre-programmed innervation (41;61). Neural activation and innervation of muscle can therefore be seen as the control mechanism by which sufficient stiffness is formed in the muscle-tendon complex to preserve the tension on the tendon during movement and to prepare the muscle for force production (61). On the basis of the findings in this study, it can be concluded that musculotendinous stiffness regulation seemingly occurs independent of the tendon stiffness measured in this study, or that the

neural regulation of stiffness supersedes the intrinsic properties of the muscle-tendon complex. What this essentially implies is that all the stiffness regulation that takes place in the muscle-tendon complex during vertical jumps arises from neural mechanisms, i.e. via central pre-programming, pre-activation and reflex facilitation.

As there is a large increase in eccentric EMG compared to concentric EMG during stretch shortening cycle movements (9), it has to be assumed that there is significantly more muscle activation during the eccentric phase of these movements. One can in turn therefore make the supposition that the significant increase in EMG shown in our results during the total contact period might therefore predominantly reflect the changes in the eccentric rather than concentric phase of stretch shortening cycle activities.

This rationale in part seems to be supported by the findings of Gollhofer et al. (58). Their EMG: force relationship when expressed for the total contact period seemed to primarily be related to the EMG: force relationship of the eccentric phase of stretch shortening cycle activities. Gollhofer et al. (58) found changes in the eccentric EMG: force relationship, which were matched by similar changes in the EMG: force relationship for the total contact period. Yet these changes were not mirrored in the concentric phase of the same stretch shortening cycle activities.

### **FORCE: EMG RATIO**

In addition, the integrated EMG, integrated force (impulse) and the F: EMG ratios of squat jumps and countermovement jumps were compared to each other. Muscle activation, represented by EMG, increased significantly by  $28 \pm 21\%$  in countermovement jumps over squat jumps. Impulse (defined by the area under the force-time curve) in a similar fashion increased significantly by  $60 \pm 9\%$ . An interesting finding, however, was that the F: EMG ratio (represented by dividing the integrated force data by the integrated EMG data) also increased significantly by  $29 \pm 26\%$  between the two conditions.

By reviewing the findings in this study it is clear that both EMG and impulse are positively affected by the stretch shortening cycle and that both these effects are associated with improved jumping performance. An interesting point however is that when examining the F: EMG ratio, force output increases substantially more than that which can be explained

by the increase in muscle activation alone demonstrated in this study. This is apparent in the additional consequent and significant increase in the F: EMG ratio, which implies that the countermovement jumps are more economical and that a certain amount of the work performed is elastic of origin (23).

As part of the stretch shortening cycle movement, the eccentric phase modifies not only the concentric force output, but also the neural and metabolic reactions and their relative contributions to the concentric performance enhancement (90). As high muscle stiffness is related to increased muscle recruitment (115), and furthermore, as stiffness has previously been shown to enhance elastic energy utilisation (27;30), it is logical to presume that an increased EMG during countermovement jumps would indicate increased muscle recruitment, -stiffness and therefore also storage and utilisation of elastic energy. This explanation is feasible as the increase in muscle stiffness transfers the majority of the stretch induced by eccentric muscle action to the more compliant tendons (67;83;85;124;128;137). The tendons store and then liberate the stored elastic energy in the ensuing concentric contraction (2;116;132).

One cannot however, exclude the possible contributions of the other components of stretch shortening cycle muscle function, which were not measured in this study, to the improved performance observed during the countermovement jumps. For example, the improved length-tension and force-velocity relations of muscle, and the force-acceleration effect (discussed in Chapter 1) may also have contributed to significantly increasing functional performance or jump height, but were not measured in this study.

## **JUMP HEIGHT AND PERFORMANCE DATA**

Another finding in this study was that the force-plate data correlated well with the functional performance data of both squat jumps and countermovement jumps. Squat jump height and countermovement jump height were significantly correlated to the average power output, peak power output, net impulse and average force measured on the custom-built force plate.

These findings are in agreement with other studies. For example, Finni et al. (51) found that power correspondingly increased with greater vertical jump heights. Kubo et al. (101) discovered a significant correlation between peak forces and jump height in both squat jumps ( $r = 0.75$ ) and countermovement jumps ( $r = 0.74$ ). Bosco and Komi (24) also found that net impulse ( $r = 0.75$ ,  $p < 0.001$ ), average force ( $r = 0.55$ ,  $p < 0.01$ ) and average power ( $r = 0.71$ ,  $p < 0.001$ ) correlated significantly with jump height in their squat jumps. Similar correlations were found between these variables and countermovement jump height. These associations were significant for all phases i.e. unweighting, eccentric and concentric phases ( $p < 0.05 - 0.001$ ).

In summary, it may be concluded that the results derived from the NAMS unit are consistent with, if not better than what has previously been noted in the literature. It would also seem, on the basis of the abovementioned relationships, that the force-plate data in this study are an accurate representation of muscle function in the lower body.

## CONCLUSION

An attempt was made in this study to isolate tendon elasticity (stiffness) in the lower body. Tendon elasticity is one of the key components of muscle function. Next, tendon elasticity was related to the associated stretch shortening cycle performance. It may be concluded that tendon elasticity, as an isolated component of muscle function, is not related to the associated stretch shortening cycle performance. Another interpretation of these data is that there are numerous factors, which contribute to muscle function that are interlinked and involved in the enhancement of performance observed in stretch shortening cycle movements. Further research should focus on an integrated analysis approach examining the combined effect of multiple systems and their interaction with each other during dynamic muscle function.



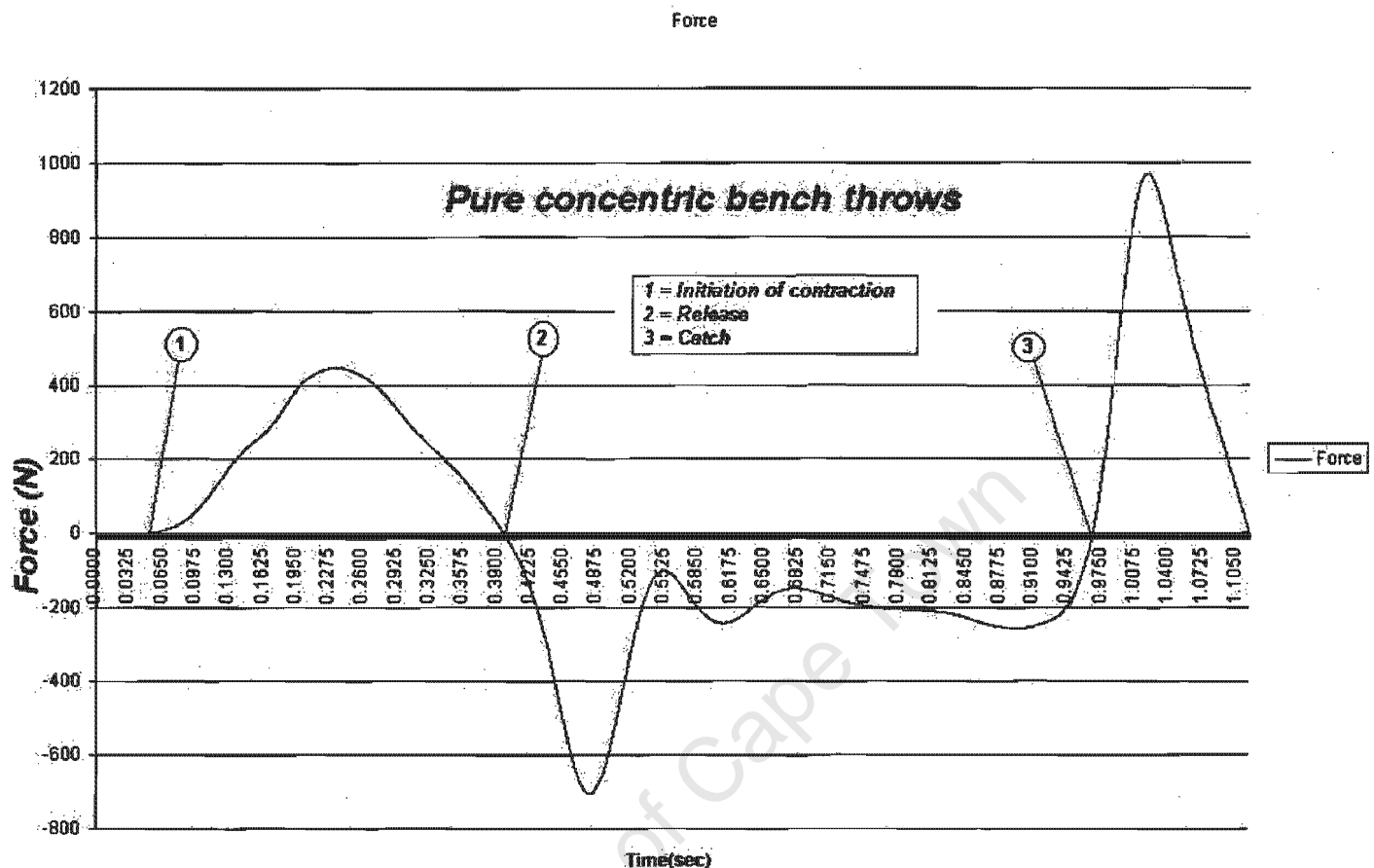
## CHAPTER 7:

# What is the relationship between tendon stiffness and stretch shortening cycle performance in the upper body?

## INTRODUCTION

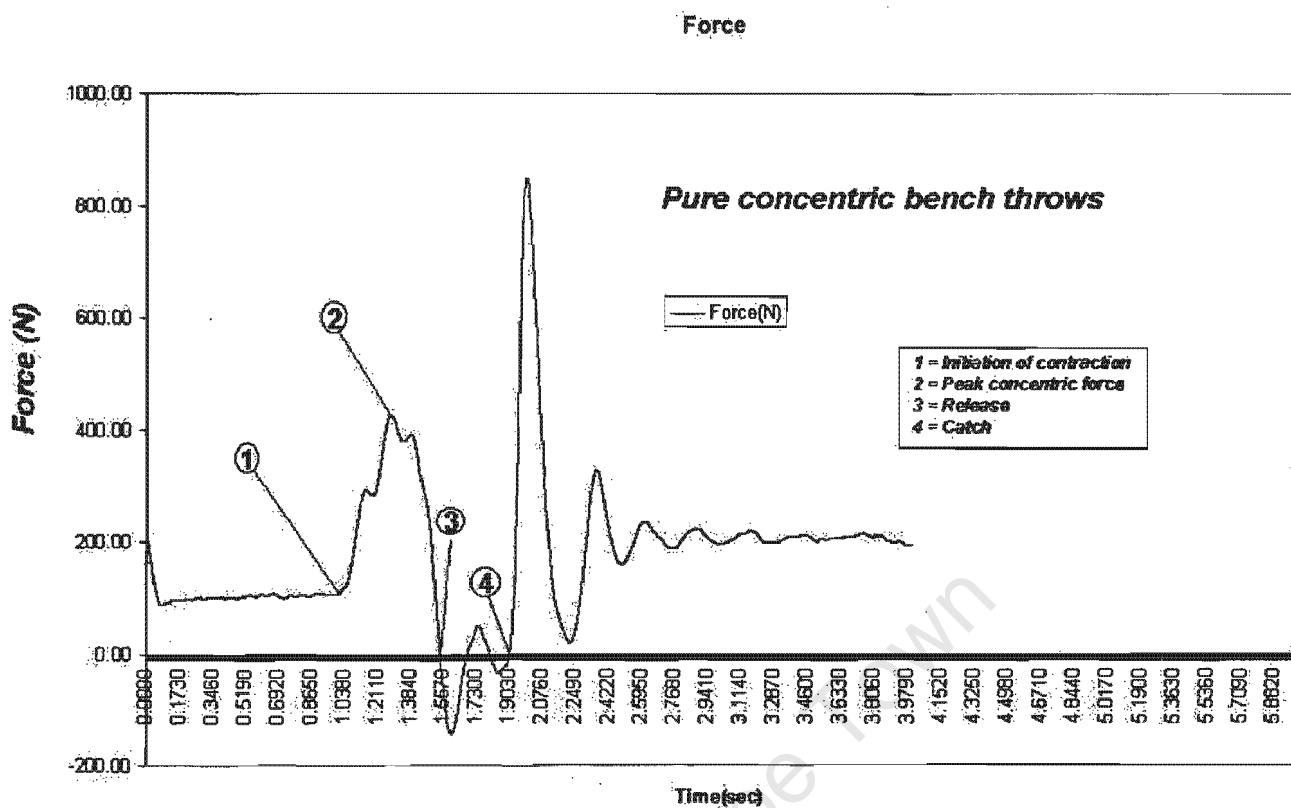
The next phase of this thesis was to relate maximal (tendon) stiffness (Chapter 3) and stretch shortening cycle ability of the upper body (Chapter 5) using the NAMS unit, to answer the following research question: *"Is there a relationship between musculotendinous stiffness and stretch shortening cycle muscle function in the upper body musculature?"* We furthermore intended to determine if similar relationships, as found in the lower body study (Chapter 6), were noted in the upper body.

As with the lower body (Chapter 6), the protocol was adapted by zeroing force-measurement of the force-plate to accommodate the weight of the subject, bolted bench press bench and bar, thereby normalizing for the combined subject-bench-bar weight (Figure 1). Figure 2 shows the original regional breakdown of pure concentric bench throws used in Chapter 5. These data were not normalized for the additional weight of the bar.



**Figure 1:** Modified force-time regional breakdown of a pure concentric bench throw after normalising the force plate to accommodate the weight of the bench, subject and bar. (\* Graph from original data)

A new regional breakdown was again defined when using the normalized force-time data. As the "Total" measures of Chapter 5 were generally again the most repeatable measurements and the "Concentric-defined" measures were generally not (especially in the upper body), the "Concentric-defined" measures were excluded from the modified regional breakdown (Figure 1).



**Figure 2:** Original force-time regional breakdown of a pure concentric bench throw (Chapter 5).  
(\* Graph from original data)

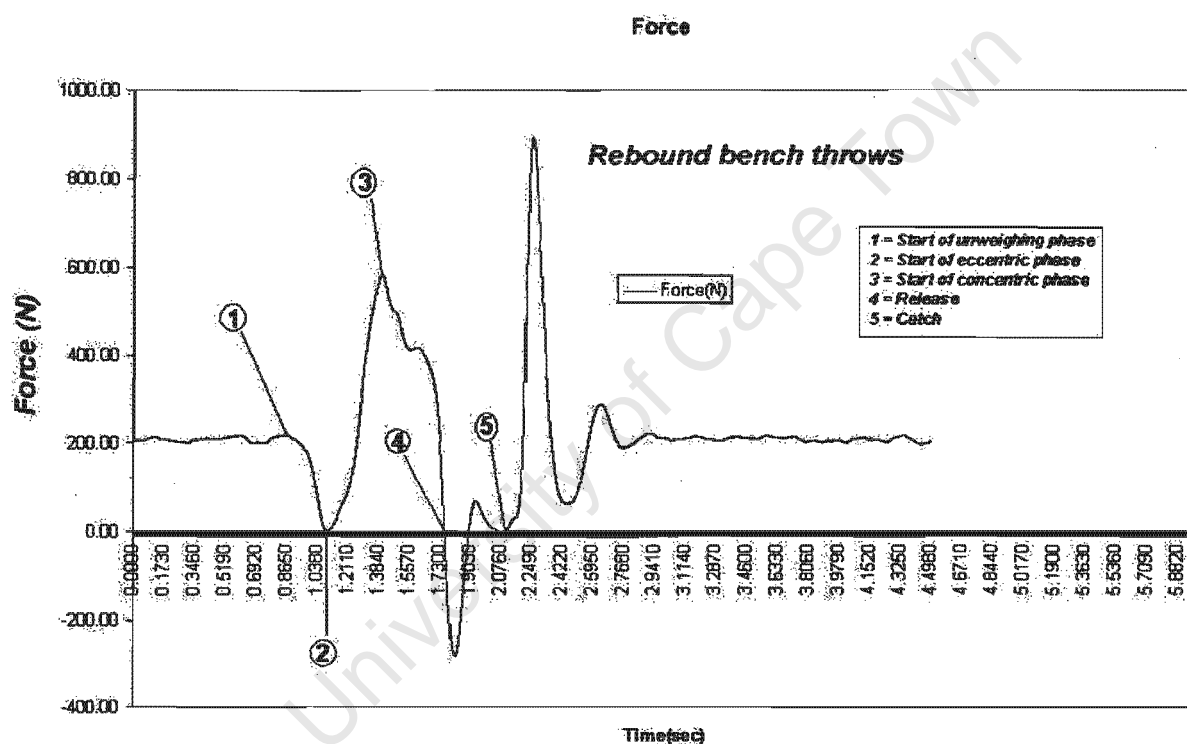
Due to oscillations of the body and bench on the force plate after release of the bar it was sometimes difficult to select the exact point where the bar was caught during the bench throws. This allowed for subjectivity and therefore human error. By zeroing the force-plate and normalizing the data for the combined subject-bench-bar weight, the determination of the modified force-time regional breakdown would be less subjective and more objective, thereby reducing the experimental error involved (Figure 1).

On this premise, as with the lower body study (Chapter 6), all original bench-throw height data (Chapter 5) were recalculated with the modified normalized breakdown to compare the repeatability results. The release (2) and catch (3) points (Figure 1) were adjusted to represent when the bar weight started to leave the subject's hands and when the full bar weight had returned to them respectively. This adjustment was treated as a systematic error as it was assumed that it was constant throughout the tests.

Pure concentric bench throw height originally had an intra-class correlation coefficient (R) of 0.99 (95% CI: 0.97-1.00) and coefficient of variation (CV) of 5.8 (95% CI: 4.4-8.6)%.

The repeatability data of the modified breakdown of pure concentric bench throws was shown to be  $R = 0.99$  (95% CI: 0.97-1.00) and  $CV = 6.3$  (95% CI: 4.8-9.3)%. Therefore, there was no change in repeatability between the two techniques i.e. the original and the modified method.

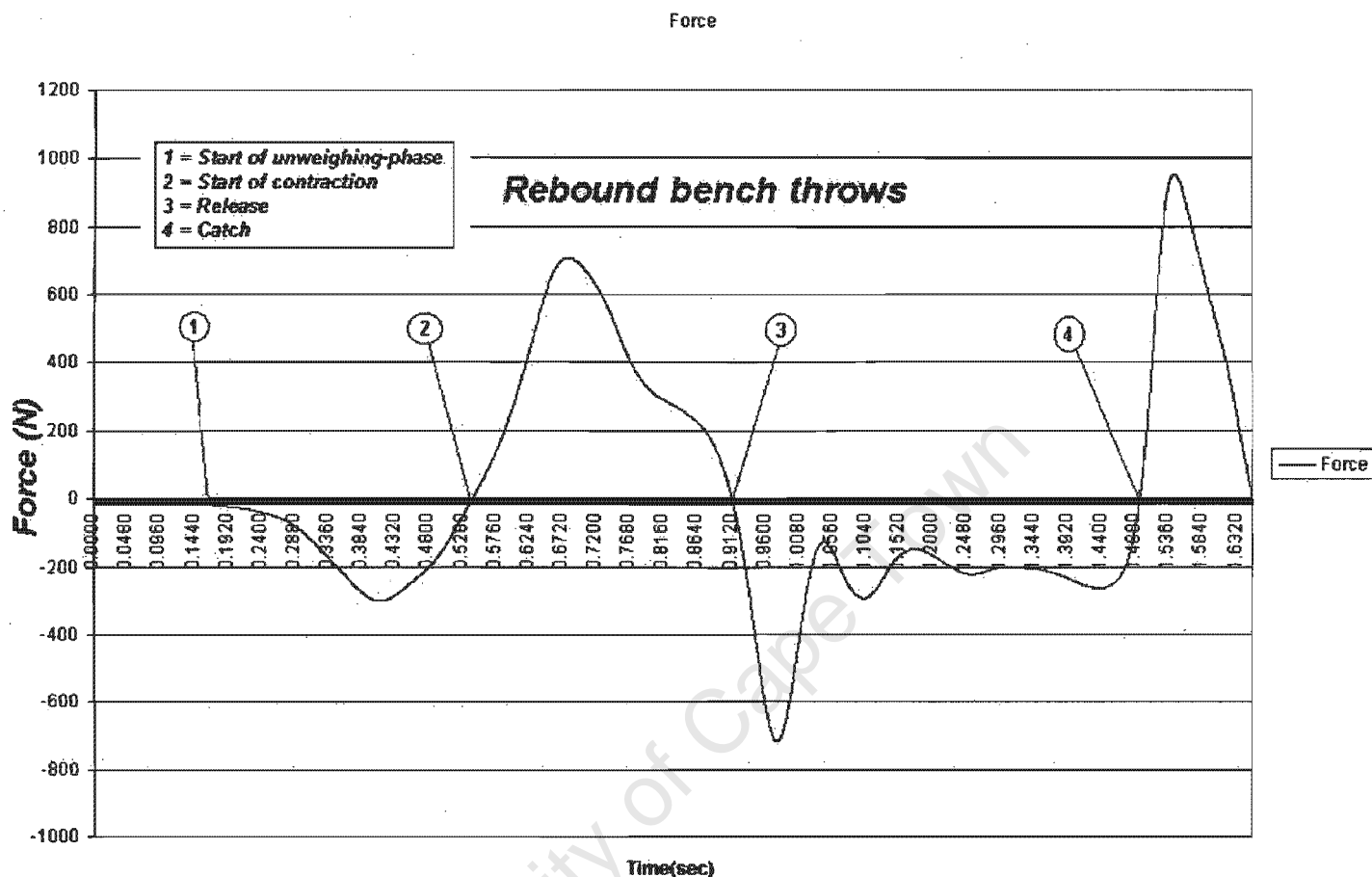
Figure 3 shows the original regional breakdown of rebound bench throws used in Chapter 5. These data were also not normalized for the combined subject-bench-bar weight. The original data, as with the pure concentric throws, only normalized for the combined bench and subject weight, and not the bar.



**Figure 3:** Original force-time regional breakdown of a rebound bench throw (Chapter 5).  
(\* Graph from original data)

As with the pure concentric bench throws, the “Total” measures of Chapter 5 were the most repeatable and the “Concentric-defined” measures were also excluded from the modified regional breakdown (Figure 4). The release (3) and catch (4) points (Figure 4), as with the pure concentric bench throws, were adjusted to represent when the bar weight started to leave the subject’s hands and when the full bar weight had returned to the subject’s hands respectively. Initiation of contraction (2) was also adjusted to when the

deceleration- or eccentric force equalled that of the bar weight. These adjustments were also treated as a systematic error as it was assumed that it was constant.



**Figure 4:** Modified force-time regional breakdown of a rebound bench throw after normalising the force plate to accommodate the weight of the bench, subject and bar. (\* Graph from original data)

Rebound bench throw height originally had an intra-class correlation coefficient  $R = 0.98$  (95% CI: 0.94-0.99) and coefficient of variation  $CV = 7.7$  (95% CI: 5.8-11.4)%. The normalized breakdown did not change the repeatability, but reduced the variability of the rebound bench throws very slightly to  $R = 0.98$  (95% CI: 0.94-0.99) and  $CV = 5.3$  (95% CI: 4.0-7.8)%.

The precision, by limits of agreement (two standard deviations above and below the average difference between trials), of the bench throws changed slightly with the range for pure concentric bench throws increasing from 8.0 cm ( $-0.3 \pm 4.0$  cm) to 10.8 cm ( $-0.6 \pm 5.4$  cm) and rebound bench throws from 9.4 cm ( $-0.4 \pm 4.7$  cm) to 12.2 cm ( $-0.7 \pm 6.1$  cm).

To compare stretch shortening cycle ability in the upper body bench throws, the bench throw ability of each subject needs to be normalised. It was therefore decided, as with the lower body study (Chapter 6), to express this relationship as a percentage change. The repeatability of the percentage bench throw height potentiation changed from  $R = 0.75$  (95% CI: 0.45-0.92) and  $CV = 71.7$  (95% CI: 54.2-106.0)%, to  $R = 0.53$  (95% CI: 0.14-0.84) and  $CV = 86.9$  (95% CI: 66.7-128.1)%. The limits of agreement also changed from 74.2% ( $0.6 \pm 37.1\%$ ) to 87.4% ( $-2.6 \pm 43.7\%$ ). Regardless of the marginally reduced repeatability of bench throw height potentiation and absence of significant improvement in repeatability of the throw height measures with the revised analysis, the modified method, because of its reduced subjectivity, was preferred over the original method used in Chapter 5.

The primary aim of this study was therefore to quantify the *in vivo* elastic properties of muscles and tendons in the upper body and to investigate their relationship with bench throw performance and the potentiation of the stretch shortening cycle during bench throws. A secondary aim of this study, accepting the limitations discussed in Chapter 5 and those mentioned above, was to investigate and attempt to understand the underlying relationships between the various measures of stretch shortening cycle ability in the upper body.

## **METHODS**

### **SUBJECTS**

31 Male subjects with varying training backgrounds and at least one year's resistance-training experience were recruited for this study. Only 19 of these subjects completed the study. Of the 12 subjects that were excluded from the study, 1 retired midway, 5 were unable to support the load with sufficient stability during the oscillation technique to be tested further, and the remaining 6 subjects could not tolerate more than 3 - 4 loads of the minimum of 5 required loads on the oscillation test. The remainder of the subjects ( $n = 19$ ) ranged between 20 and 32 years old.

All subjects were informed about the nature and inherent risks involved in the study. The Ethics Committee of the University of Cape Town approved this study. Each subject

completed a questionnaire regarding their medical, training and general information and further signed an informed consent form complying with the guidelines set by the American College of Sports Medicine (ACSM) (3), before participating in this study. The general characteristics of the subject group are summarized in Table 1. Body fat was expressed as a percentage (43) and sum of 7 skinfolds (7).

**Table 1:** General characteristics of the subject group (n = 19).

AGE (yrs)	HEIGHT (cm)	MASS (kg)	SUM OF SKINFOLDS (mm)	% BODY FAT	LEAN BODY MASS (kg)
24.9 ± 3.0	179.5 ± 7.5	81.9 ± 9.9	76.3 ± 33.4	14.9 ± 4.4	69.7 ± 8.6

**EXPERIMENTAL DESIGN**

The trial consisted of testing on three days, each separated by at least 48 hours. On each day, the testing was preceded by a standardized warm-up consisting of 5 minutes of continuous, self-paced, submaximal cycling on a stationary bike, 2 sets of 10 push-ups with body mass as resistance, specific stretches for the upper body and 1 set of 10 repetitions bench press on the NAMS unit with 21.32 kg (zero load on the bar).

The first day was used as a familiarization trial to teach the various techniques to the subjects and familiarize them with the procedures involved in the study. Earlier studies (Chapter 3) have shown that certain individuals have difficulty in acquiring the oscillation technique and a significant amount of time was spent in familiarizing them with this test. This was also the case with the lower body test (Chapters 3, 4 and 6). On the basis of the findings in Chapter 4 using the lower body, the oscillation test in the upper body was also modified to an absolute incremental loading protocol.

Subjects were first familiarized with the modified oscillation test procedure, as this was the test that would determine whether they were allowed to continue with the remainder of the study. They were given 4 - 6 trials per load to familiarize themselves and the loads were increased incrementally to just below their expected maximum tolerable load.

If the subject managed the oscillation technique, he was further familiarized with the remainder of the testing procedures i.e. the MVC bench press and bench throw tests. Subjects were given 3 - 6 trials per test to familiarize themselves with the techniques. After the familiarization day, the subjects had to visit the laboratory on two more occasions.

On the second day body composition assessment and the modified upper body oscillation test were assessed. On the third testing day, the subject performed the MVC bench press and bench throw tests.

To control for factors that might affect muscle function adversely, subjects were asked to refrain from any heavy or intense training during the testing period and also to refrain from any upper body exercise at least 24 hours before each testing day. Subjects had at least 48 hours rest between the familiarisation and both testing days.

#### **THE MODIFIED OSCILLATION TEST FOR THE UPPER BODY**

For the modified upper body oscillation test a similar procedure to that described in Chapter 3 (p 153) was used. Subjects were instructed to maintain a constant muscular activity and force and not to respond to the perturbation in any way (156). The subjects were, however, not tested at relative loads calculated as a percentage of their maximum rebound bench press as were used in the original studies (Chapter 3). Subjects were tested at the same absolute loads in a way similar to that used in the modified lower body protocol discussed in Chapters 4 and 6.

The initial load used was that of the bar alone i.e. 21.32 kg. Subjects were further tested at loads increasing in 7.5 kg increments until they could no longer maintain the load with sufficient stability to enable accurate measurement of the force-oscillations. The calculated stiffness at the maximum load, i.e. where the force data became unstable, was compared to the rest of the data. If this value was a clear outlier, the data for this load was excluded from the ensuing calculation of maximal stiffness. The decision to exclude this data point was confirmed by an impartial reviewer. The test on average required the subjects to be tested at approximately 5 - 8 loads.



The force data were sampled at a frequency of 2000 Hz using LabVIEW version 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were then used for further analysis.

The analysis of the resultant force oscillations and calculation of elastic stiffness of the muscle-tendon complex used the same formulation and procedures previously outlined in Chapter 4 for the lower body (p 169 - 170).

#### **TESTING FOR THE MVC ISOMETRIC BENCH PRESS**

The same set-up and procedure for the MVC isometric bench press test was used as previously described in Chapter 5 (p 202 - 203).

#### **THE SSC TESTS FOR THE UPPER BODY**

The same techniques and controls as discussed in Chapters 2 (p 126 - 127) and 5 (p 203) were used for the testing of bench throws i.e. pure concentric bench throws and rebound bench throws. One additional modification was made to the pure concentric bench throw test. In an attempt to make the test more sensitive to differences, subjects were instructed to rest the bar on the chest and start the throw from a completely relaxed position rather than from a suspended position  $\pm 3$  cm above the chest.

All the force data, as for the MVC bench press, were sampled at a frequency of 2000 Hz using LabVIEW 6.0.2 software (National Instruments, Austin, Texas, U.S.A.) and passed through a second order lowpass Butterworth filter with a cut-off frequency of 7 Hz. The filtered force data were used for further analysis.

For the analysis of force and EMG, the data were segmented using the modified force-time regional breakdown mentioned earlier in this chapter.

For the force- and EMG measures of pure concentric bench throws (Figure 1), the data between initiation of contraction (1) and release (2) were used. The flight time, which was calculated from the time difference between release (2) and catch (3) was used for calculation of pure concentric bench throw height.

The data between start of contraction (2) and release (3) were used for the force- and EMG measures of rebound bench throws (Figure 4). For calculation of rebound bench throw height, the flight time, which was calculated from the time difference between release (3) and catch (4) was used.

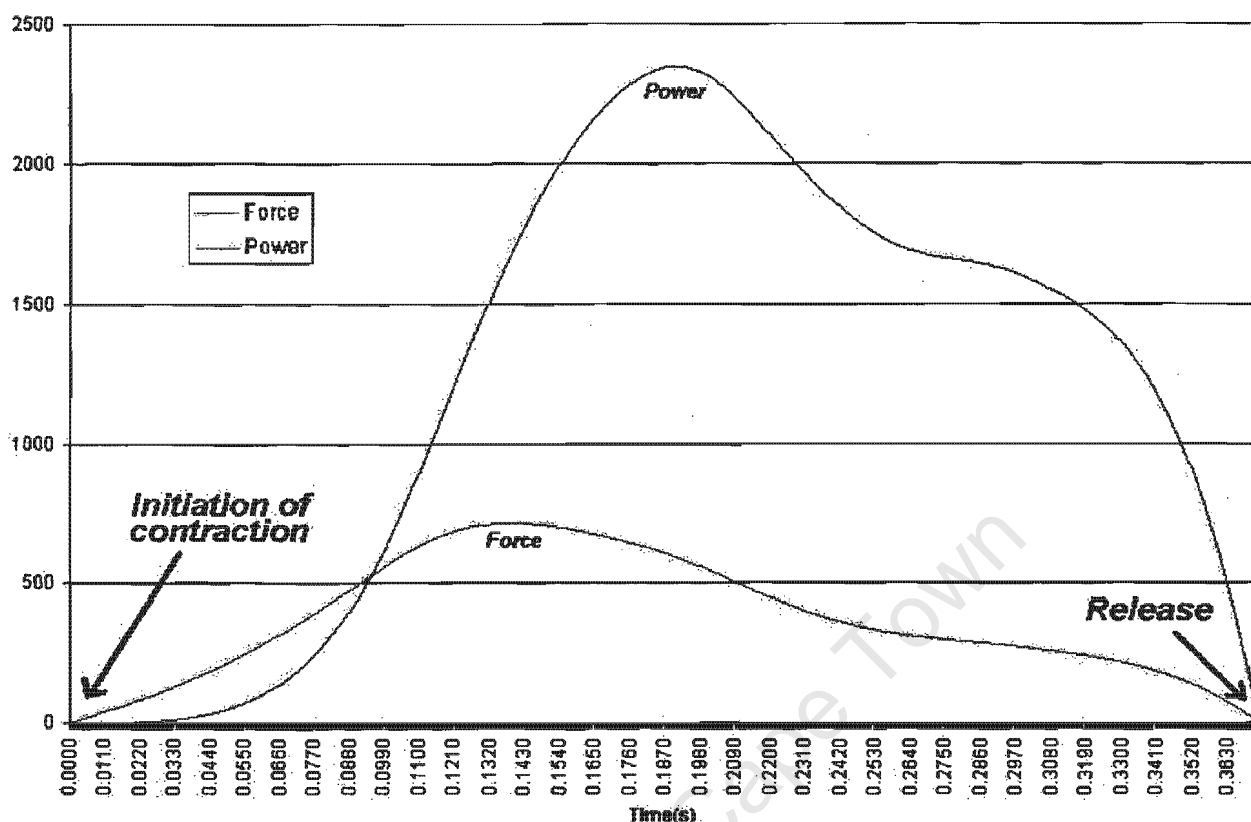
All force and EMG data were analysed using this regional breakdown. The same formulae used in Chapter 5 (p 205) were again used in the calculation of throw height for both tests.

As with the previous lower body study (Chapter 6), instantaneous power, using the abovementioned regional breakdown and the formulae of Harman et al. (69) were calculated in an attempt to have an alternative analysis of bench throw data and stretch shortening cycle potentiation.

The same procedure, as previously discussed in Chapter 6 (p 232), was used for calculation of instantaneous power (Figure 5). The only difference was that the equation to calculate vertical velocity was changed to:

$$\Delta V_v \text{ (m.s}^{-1}\text{)} = \{\text{Vertical ground reaction force (N) x time period (s)}\} / \text{bar mass (kg)}$$

(i.e. subject mass was not included because only the bar mass was moved.)



**Figure 5:** The calculation of instantaneous power is calculated throughout the bench throw test using the modified regional breakdown. (\* Graph from original data)

### ELECTROMYOGRAPHY (EMG)

For the modified upper body oscillation test, surface EMG was measured to identify trials, which were confounded by neurological intervention (149;151;155;156). The same procedure was used as described in Chapters 3 (p 153 - 154).

For the bench throws the same sampling strategy, preparation, muscle group and procedure was used for EMG analysis, as previously described in Chapter 5 (p 205 - 206). The only modifications to the analysis were the alterations to the regional breakdown mentioned earlier (Figures 1 and 4).

It must be noted that both upper limbs contributed towards the force generated in the trials whereas the EMG was only measured on the right upper limb only. However, as with vertical jumps in the lower body, bench throws were assumed to be a bilaterally

symmetrical movement. We therefore selected the right limb as being representative of both sides. Because rebound throws were compared to pure concentric throws in the same individuals, the limited error in this assumption was acceptable.

## STATISTICAL ANALYSIS

The same statistical procedures were used as previously described in Chapter 6 (p 234).

## RESULTS

### MUSCULOTENDINOUS STIFFNESS VS BENCH THROW DATA

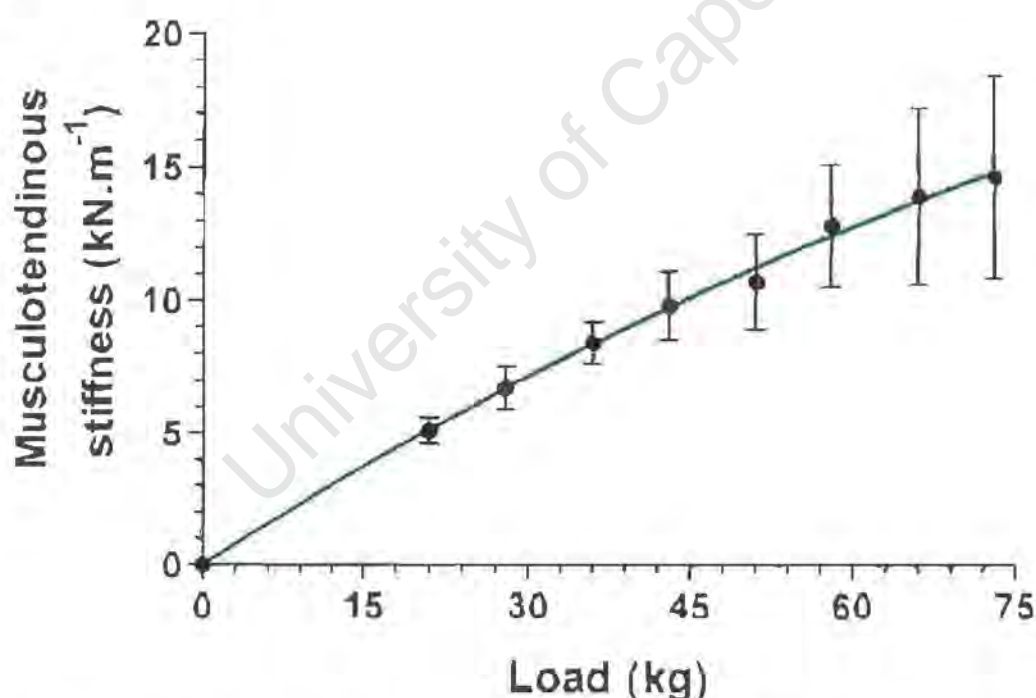
The range of musculotendinous stiffness values, maximal (tendon) and submaximal, of the subjects tested ( $n = 19$ ) in this study using the modified oscillation technique are represented in Table 2. The tendon stiffness values ranged from 8.3 - 35.2  $\text{kN.m}^{-1}$  averaging  $15.6 \pm 7.3 \text{ kN.m}^{-1}$ . The subjects managed on average between 5 - 6 loads before the data became unstable.

**Table 2:** Musculotendinous stiffness values ( $\text{kN.m}^{-1}$ ) of the subjects over the range of absolute loads and the maximal predicted stiffness of the subjects ( $n = 19$ ).

Subject	21.3 kg	28.8 kg	36.3 kg	43.8 kg	51.3 kg	58.8 kg	66.3 kg	73.8 kg	81.3 kg	88.8 kg	96.3 kg	Maximal stiffness
Up1	4.9	6.5	9.0	11.7	13.9	14.8	16.1	16.8	16.8	-	-	17.7
Up2	4.8	6.1	8.0	10.0	11.1	14.0	12.9	13.8	14.3	15.0	-	15.3
Up3	4.7	7.2	8.7	10.0	10.8	-	-	-	-	-	-	11.5
Up4	6.1	8.2	9.8	10.0	10.4	11.2	11.5	12.5	-	-	-	13.0
Up5	4.3	6.0	8.0	10.0	11.1	12.3	12.0	12.5	13.3	-	-	13.2
Up6	5.2	7.5	8.7	11.7	12.5	-	-	-	-	-	-	17.5
Up7	5.2	7.0	8.7	9.2	9.8	-	-	-	-	-	-	10.2
Up8	5.1	6.9	8.8	10.5	11.1	12.8	-	-	-	-	-	18.0
Up9	5.1	6.4	8.3	10.1	11.6	13.3	13.4	-	-	-	-	18.5
Up10	4.3	4.8	6.6	7.2	8.2	-	-	-	-	-	-	15.2
Up12	5.4	7.0	9.3	12.4	13.6	15.1	18.6	17.7	19.6	18.9	-	20.7
Up13	5.5	7.6	9.1	9.7	12.6	13.5	-	-	-	-	-	35.2
Up15	5.4	6.8	8.4	9.6	9.7	-	-	-	-	-	-	10.7
Up16	5.4	6.5	8.0	10.3	11.9	15.6	17.8	20.0	20.3	22.4	24.8	32.2
Up17	5.8	7.6	7.9	8.6	9.7	8.4	-	-	-	-	-	9.0
Up18	5.2	7.2	9.0	9.6	11.2	-	-	-	-	-	-	18.0
Up19	5.1	6.3	8.5	8.5	8.0	-	-	-	-	-	-	8.5
Up22	5.6	7.2	8.0	9.1	8.3	-	-	-	-	-	-	8.3
Up24	4.5	5.3	6.5	7.5	8.3	9.4	9.1	8.8	-	-	-	9.6
Average	5.1	6.7	8.4	9.8	10.7	12.8	13.9	14.6	16.9	18.8	24.8	15.6
± SD	± 0.5	± 0.8	± 0.8	± 1.3	± 1.8	± 2.3	± 3.3	± 3.8	± 3.1	± 3.7		± 7.3

Figure 6 illustrates the grouped average profile of the stiffness-load relationship of the subjects used in this trial. The three higher loads were omitted from this graph, as there were too few subjects that managed these loads with sufficient stability. Only 5 subjects managed the 81.3 kg load, 3 subjects the 88.8 kg and 1 the 96.3 kg load. It became more difficult for the subjects to maintain the loads as they became heavier, irrespective of the strength of the subject.

The majority of the stiffness data followed the classical curvilinear stiffness-load relationship shown in the earlier studies (Chapters 3, 4, and 6). However not all subjects could maintain loads heavy enough to visually induce the formation of a plateau. This was systematically corrected for, as all data were fitted to the Boltzmann equation and extrapolated to a maximum plateau. Musculotendinous stiffness values, observed in the individual subject data, became more variable as the loads were increased. Based on previous data available from earlier studies (Chapter 3) and previous research (156) this was an expected finding.



**Figure 6:** The grouped average stiffness-load relationship data of the subjects used in this study ( $n = 19$ ). Only the data up to 73.8 kg were plotted as very few subjects managed heavier loads.

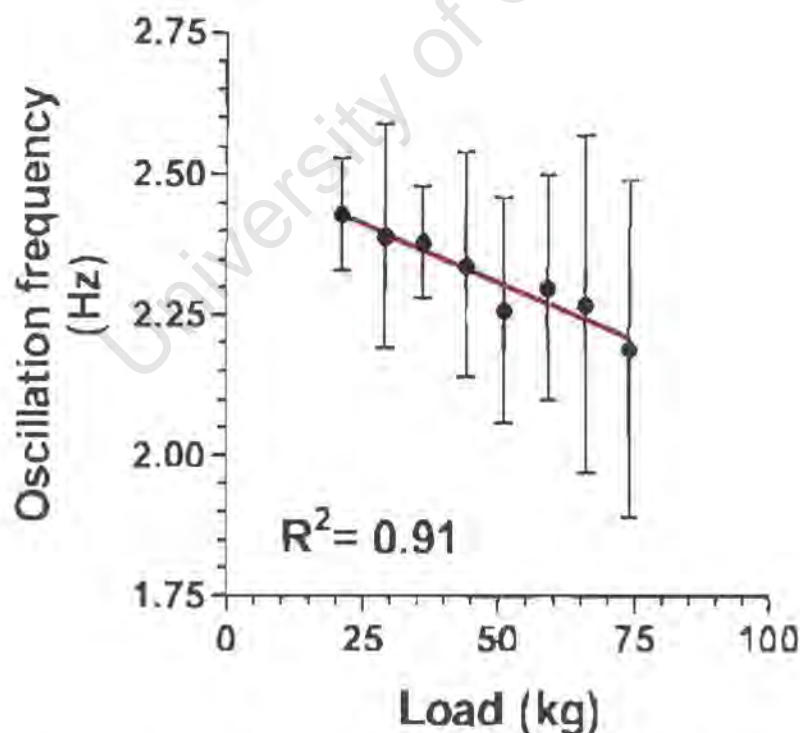
The relationship between the damped natural frequency of oscillation and increasing load followed the same pattern that we had found in our earlier studies (Chapters 3, 5 and 6). The oscillation frequency decreased, as the load on the muscle-tendon complex was

increased. This relationship (Figure 7) was however far more variable than in the lower body study (Chapter 6), but nevertheless followed the same tendency. As mentioned earlier, there were insufficient subjects tested at the 81.3 - 96.3 kg loads. The data from these loads were therefore excluded in Figure 7 and Table 3.

Table 3 shows that the damped natural frequency of oscillations in the upper body trial ranged from 2.19 - 2.43 Hz over the various loading conditions.

**Table 3:** The average upper body damped natural frequency data at the relative loads using the NAMS unit (n x 6 trials per load).

Absolute load (kg)	Damped Natural Frequency (Hz)
21.3	2.43 ± 0.13
28.8	2.39 ± 0.16
36.3	2.38 ± 0.13
43.8	2.34 ± 0.17
51.3	2.26 ± 0.20
58.8	2.30 ± 0.22
66.3	2.27 ± 0.26
73.8	2.19 ± 0.29



**Figure 7:** The response of the damped natural frequency of oscillation of the muscle-tendon complex under incremental loading conditions (n = 19).



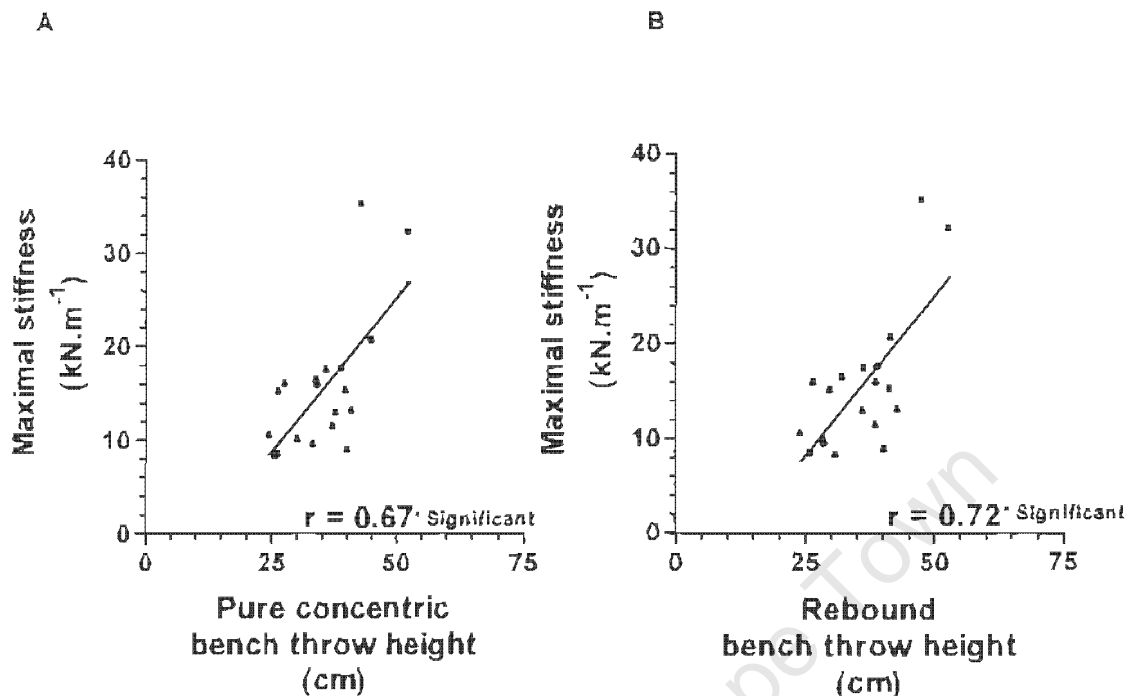
Table 4 presents the range of data, which was used in the calculation of stretch shortening cycle potentiation in the bench throw tests in the upper body.

**Table 4:** Data collected during the bench throw tests used in calculation of stretch shortening cycle performance potentiation ( $n = 19$ ).

Subject	PCBT (cm)	RBT (cm)	PCBT Impulse (Ns)	RBT Impulse (Ns)	PCBT IEMG (%MVC)	RBT IEMG (%MVC)	PCBT F: EMG (N/%)	RBT F: EMG (N/%)	PCBT AP (W)	RBT AP (W)	PCBT PP (W)	RBT PP (W)
Up1	38.8	38.9	92.4	132.6	10.0	10.9	9.2	12.1	628	1154	1510	2457
Up2	39.5	41.3	84.5	139.3	9.1	10.8	9.3	12.9	422	1240	1100	2348
Up3	37.0	38.5	73.9	132.6	14.7	16.2	5.0	8.2	327	1058	769	2010
Up4	37.7	36.1	71.2	122.8	11.1	11.7	6.4	10.5	342	871	693	1975
Up5	40.9	42.8	85.5	144.6	10.8	12.2	7.9	11.8	521	1300	1102	2332
Up6	35.8	36.2	73.1	122.7	13.7	16.1	5.3	7.6	338	884	716	1356
Up7	30.0	28.3	65.5	107.6	14.9	18.2	4.4	5.9	283	669	500	1069
Up8	27.6	26.5	68.7	121.2	15.8	16.9	4.3	7.2	309	879	669	1823
Up9	33.7	32.1	69.4	117.9	7.7	10.8	9.0	10.9	422	857	808	1868
Up10	26.3	29.7	64.8	93.8	16.1	14.6	4.0	6.4	226	477	733	825
Up12	44.8	41.4	105.1	156.3	9.4	7.8	11.2	20.0	715	1976	1787	3909
Up13	42.8	47.4	122.6	184.8	14.8	16.6	8.3	11.1	1031	2139	2521	4798
Up15	24.5	24.1	62.2	105.5	13.3	15.1	4.7	7.0	225	667	462	966
Up16	52.3	52.7	107.9	161.0	7.3	9.3	14.8	17.4	849	1720	2170	3603
Up17	39.9	40.1	93.6	153.7	8.7	10.1	10.8	15.2	623	1545	1264	2388
Up18	34.0	38.6	81.5	130.1	13.0	16.8	6.2	7.7	385	1137	974	1995
Up19	26.2	25.8	67.3	116.5	9.0	10.3	7.5	11.3	278	736	529	1456
Up22	25.7	30.7	64.4	115.2	19.5	15.3	3.3	7.5	227	875	473	1688
Up24	33.1	28.6	71.0	111.0	9.1	9.9	7.8	11.2	324	687	801	1820
Average ± SD	35.3 ± 7.4	35.8 ± 7.8	80.2 ± 17.1	130.0 ± 22.4	12.0 ± 3.4	13.1 ± 3.2	7.3 ± 3.0	10.6 ± 3.8	446 ± 226	1099 ± 461	1031 ± 585	2141 ± 1013

**Index:** PCBT = pure concentric bench throws, RBT = rebound bench throws, IEMG = integrated EMG, F: EMG = Force to EMG ratio, AP = average power, PP = peak power

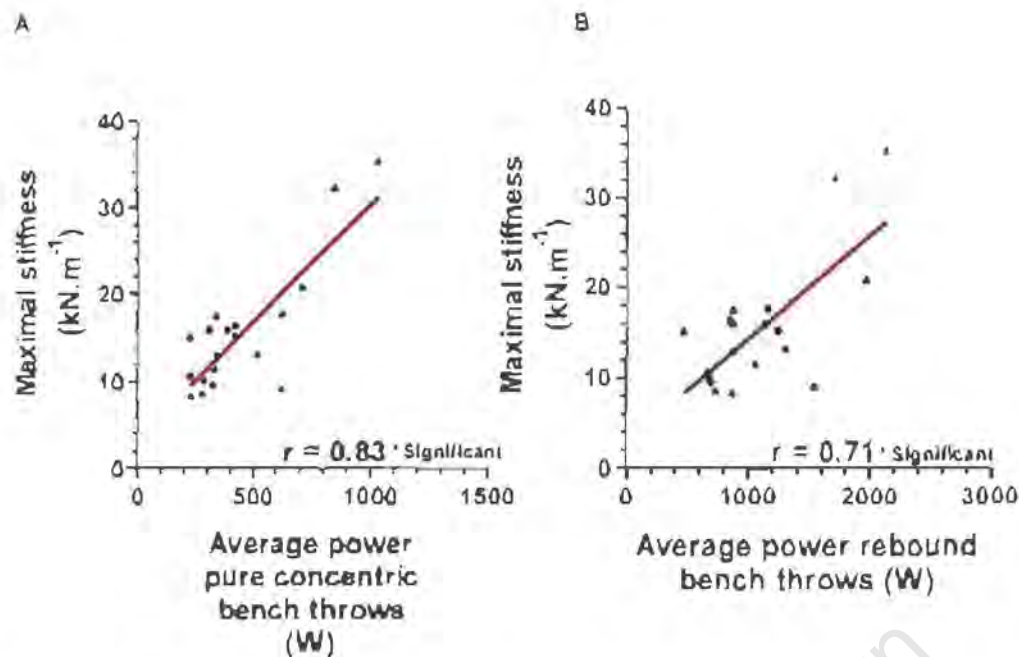
Figure 8 shows that there were significant relationships between maximal (tendon) stiffness and both (A) pure concentric bench throw height ( $r = 0.67$ ,  $p < 0.002$ ) and (B) rebound bench throw height ( $r = 0.72$ ,  $p < 0.0006$ ). These relationships were however not very good, as tendon stiffness could only account for 45% and 51% of the variability of pure concentric bench throws and rebound bench throws respectively.



**Figure 8:** A scatterplot of maximal (tendon) stiffness and bench throw height in pure concentric bench throws (A) and rebound bench throws (B) ( $n = 19$ ).

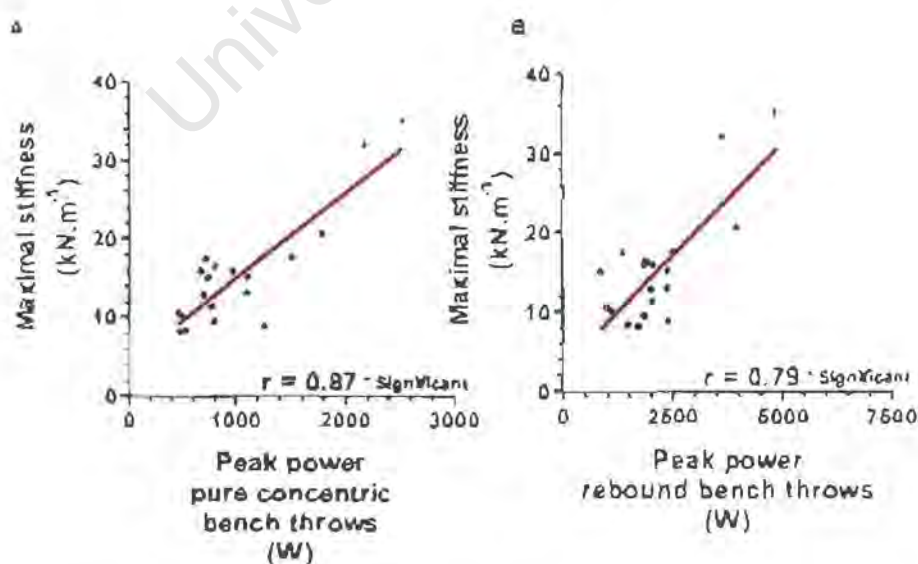
Figure 9 indicates that there were significant relationships between tendon stiffness and average power output during both (A) pure concentric bench throws ( $r = 0.83$ ,  $p < 0.0001$ ) and (B) rebound bench throws ( $r = 0.71$ ,  $p < 0.0006$ ). There was a good relationship between tendon stiffness and average power in pure concentric bench throws, as tendon stiffness could account for at least 68% of the variability in average power. The relationship between tendon stiffness and rebound bench throw average power however, was not as good as it could only account for 51% of the variability in average power.





**Figure 9:** A scatterplot between maximal (tendon) stiffness and average power generated in pure concentric bench throws (A) and rebound bench throws (B) ( $n = 19$ ).

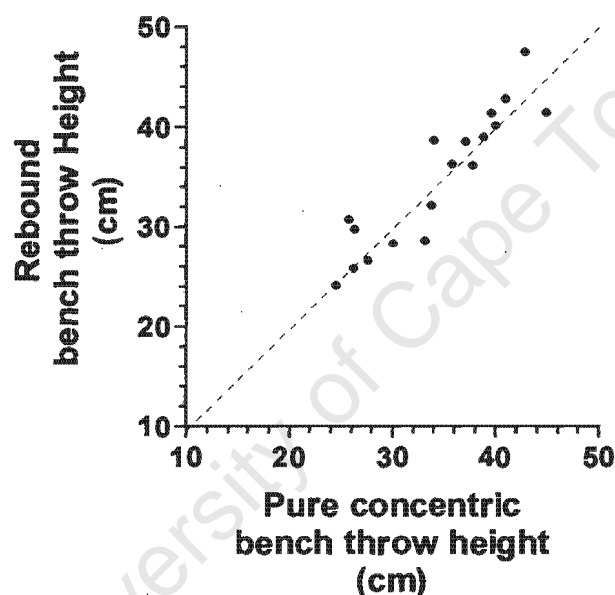
Figure 10 shows a similar pattern to that of Figure 9, however the relationships between tendon stiffness and peak power in pure concentric bench throws ( $r = 0.87$ ,  $p < 0.0001$ ) and rebound bench throws ( $r = 0.79$ ,  $p < 0.0001$ ) were better. There was a good relationship between tendon stiffness and peak power in both bench throws, as tendon stiffness could account for at least 76% and 62% of the variability in peak power for pure concentric- and rebound bench throws respectively.



**Figure 10:** A scatterplot between maximal (tendon) stiffness and peak power generated in pure concentric bench throws (A) and rebound bench throws (B) ( $n = 19$ ).

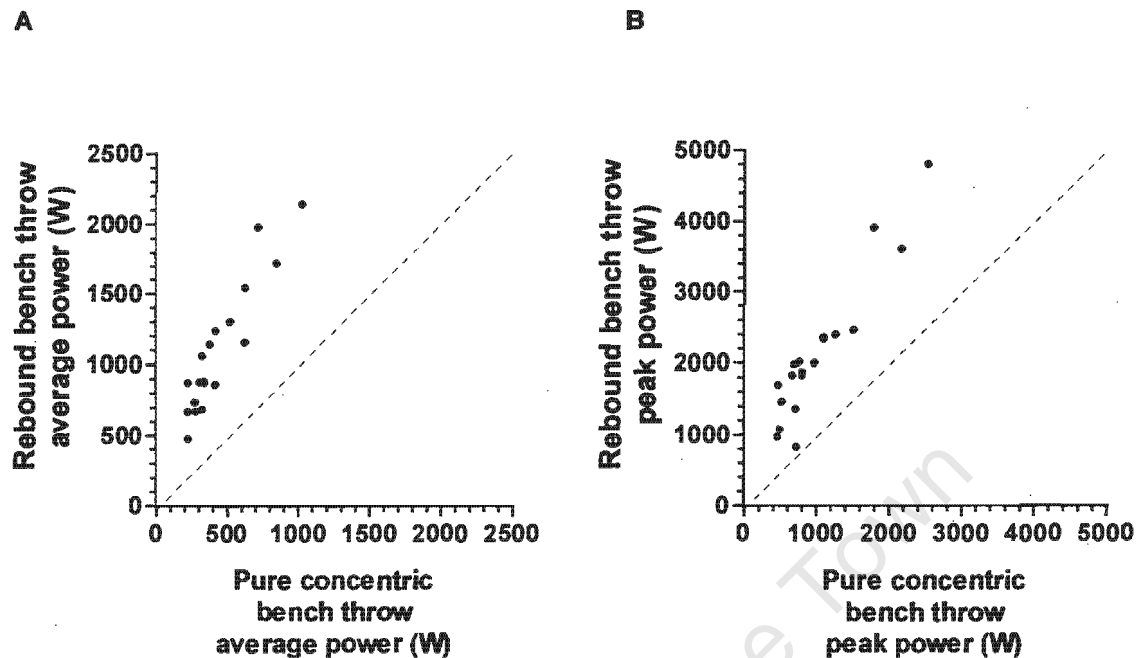
## **STRETCH SHORTENING CYCLE POTENTIATION OF BENCH THROW DATA**

Figure 11 shows the relationship between pure concentric bench throw height and rebound bench throw height. There was no significant difference ( $p < 0.4$ ) in throw height between rebound bench throws and pure concentric bench throws. No significant stretch shortening cycle improvement in bench throw height was found.



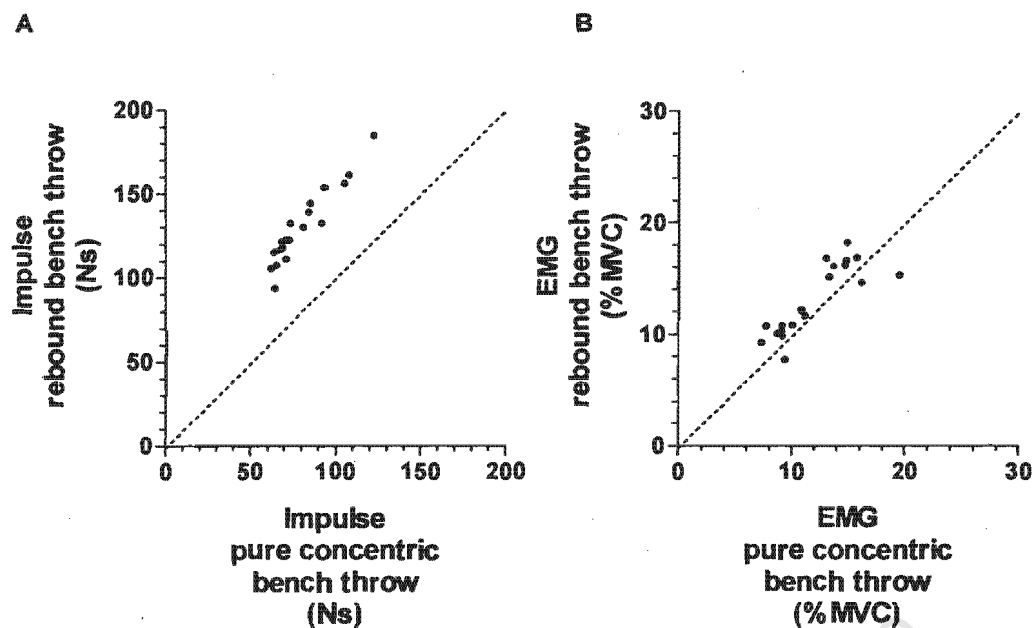
**Figure 11:** The comparison between the bench throw height in pure concentric bench throws and rebound bench throws. The data points above the line of equality (dotted line) indicate stretch shortening cycle potentiation ( $n = 19$ ).

Figure 12 compares average power (A) and peak power (B) of pure concentric bench throws to those of rebound bench throws. It was found that both the average power ( $p < 0.0000001$ ) and peak power ( $p < 0.0000001$ ) were significantly higher in rebound bench throws than in pure concentric bench throws. This indicates significant stretch shortening cycle potentiation of both average- and peak power.



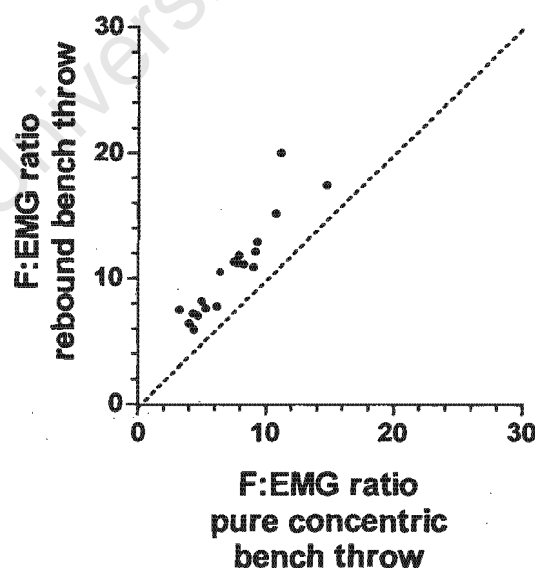
**Figure 12:** The comparison between average power (A) and peak power (B) generated in pure concentric bench throws and rebound bench throws. The data points above the line of equality (dotted line) indicate stretch shortening cycle potentiation ( $n = 19$ ).

Figure 13 shows the relationship between (A) impulse (defined by the area under the force-time curve) and (B) EMG in rebound bench throws compared to pure concentric bench throws. Both impulse ( $p < 0.0000001$ ) and EMG ( $p < 0.02$ ) were significantly increased in rebound throws versus pure concentric throws. This indicates significant stretch shortening cycle potentiation of both these measures. Average force, which relates to impulse, followed the same pattern and was significantly higher ( $p < 0.0000001$ ) in rebound throws ( $343 \pm 84$  N) than pure concentric throws ( $208 \pm 62$  N).



**Figure 13:** The comparison between impulse (A) and EMG (B) in pure concentric bench throws and rebound bench throws. The data points above the line of equality (dotted line) indicate stretch shortening cycle potentiation ( $n = 19$ ). (\* 3 Subjects in (B) did not show any potentiation using these data)

Figure 14 compares the F: EMG ratio in rebound bench throws to the F: EMG ratio in pure concentric bench throws. The F: EMG ratio increased significantly ( $p < 0.0000001$ ) in the rebound bench throws. This once again demonstrates a significant stretch shortening cycle potentiation of this measure.



**Figure 14:** The comparison between F: EMG ratios in pure concentric bench throws and rebound bench throws. The data points above the line of equality (dotted line) indicate stretch shortening cycle potentiation ( $n = 19$ ).

The stretch shortening cycle potentiation measures determined from the bench throw data are shown in Table 5.

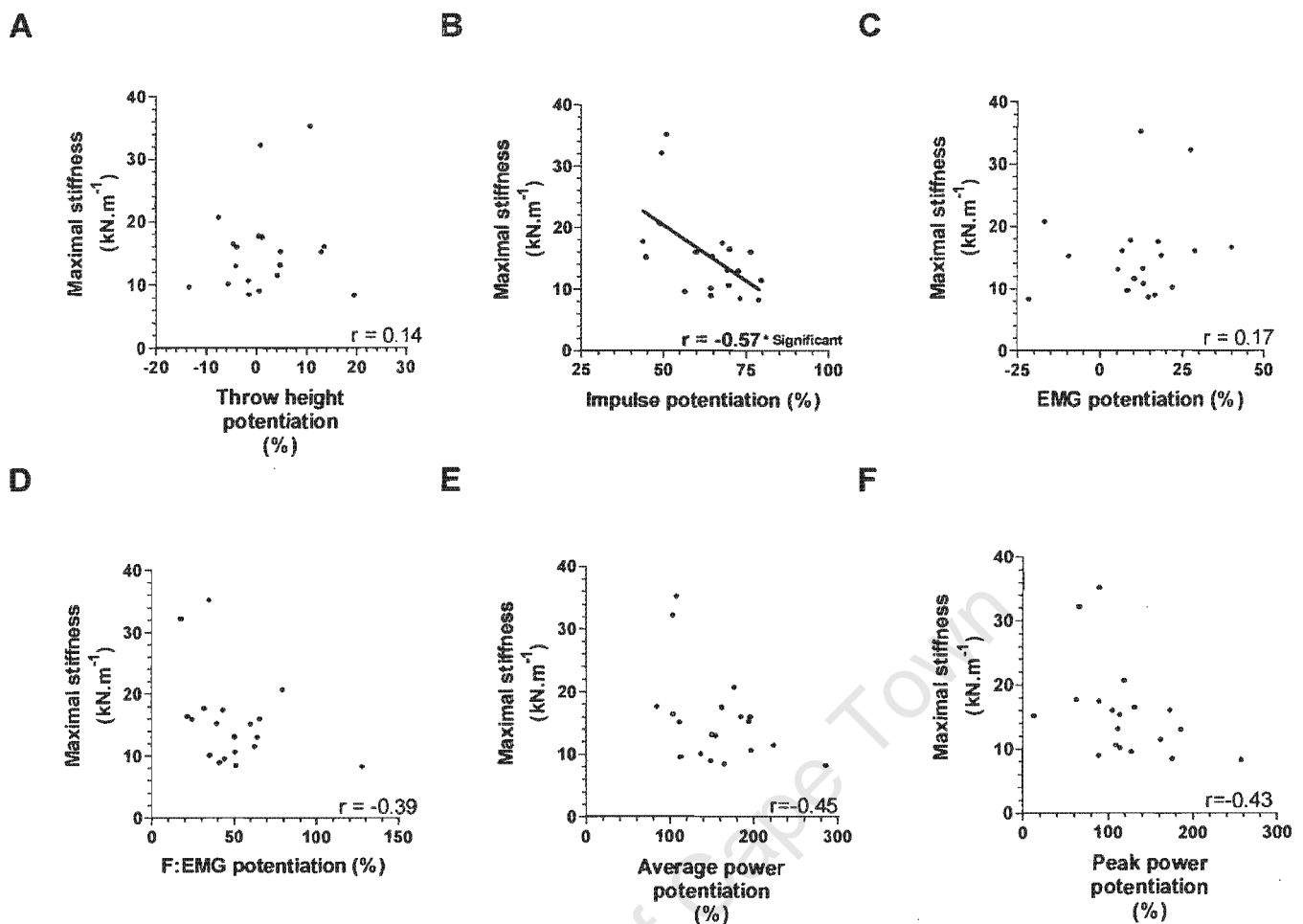
**Table 5:** *Stretch shortening cycle potentiation measures used in bench throw testing (n = 19).*

Subject	RBT-PCBT throw height Potentiation (%)	RBT-PCBT Impulse Potentiation (%)	RBT-PCBT %IEMG Potentiation (%)	RBT-PCBT F: EMG Potentiation (%)	RBT-PCBT AP Potentiation (%)	RBT-PCBT PP Potentiation (%)
Up1	0.3	43.6	9.3	31.4	83.8	62.7
Up2	4.6	64.8	18.5	39.1	193.8	113.5
Up3	4.1	79.4	10.3	62.6	223.5	161.4
Up4	-4.2	72.5	5.4	63.7	154.7	185.0
Up5	4.6	69.2	12.9	49.9	149.5	111.6
Up6	1.1	67.8	17.5	42.8	161.5	89.4
Up7	-5.7	64.2	21.8	34.8	136.4	113.8
Up8	-4.0	76.3	6.7	65.2	184.5	172.5
Up9	-4.7	69.9	39.9	21.4	103.1	131.2
Up10	12.9	44.6	-9.4	59.6	111.1	12.6
Up12	-7.6	48.7	-16.8	78.8	176.4	118.7
Up13	10.7	50.7	12.3	34.2	107.5	90.3
Up15	-1.6	69.7	13.2	49.9	196.4	109.1
Up16	0.8	49.2	27.3	17.3	102.6	66.0
Up17	0.5	64.2	16.6	40.9	148.0	88.9
Up18	13.5	59.7	28.7	24.1	195.3	104.8
Up19	-1.5	73.1	14.7	50.9	164.7	175.2
Up22	19.5	78.9	-21.5	127.8	285.5	256.9
Up24	-13.6	56.3	8.3	44.2	112.0	127.2
Average ± SD	1.6 ± 8.1	63.3 ± 11.5	11.4 ± 14.9	49.4 ± 25.0	157.4 ± 49.9	120.6 ± 53.6

**Index:** PCBT = pure concentric bench throws, RBT = rebound bench throws, IEMG = integrated EMG, F: EMG = Force to EMG ratio, AP = average power, PP = peak power

## **MUSCULOTENDINOUS STIFFNESS VS STRETCH SHORTENING CYCLE POTENTIATION**

Figure 15 shows the scatterplots between tendon stiffness and the various potentiation measures used in calculating stretch shortening cycle potentiation (Table 5) in bench throws. No significant relationships were found between tendon stiffness and (A) throw height potentiation ( $r = 0.14$ ), (C) EMG potentiation ( $r = 0.17$ ), (D) F: EMG potentiation ( $r = -0.39$ ), (E) average power potentiation ( $r = -0.45$ ) and (F) peak power potentiation ( $r = -0.43$ ). There was however a significant negative relationship between maximal (tendon) stiffness and (B) impulse potentiation ( $r = -0.57$ ,  $p < 0.01$ ). This could however be considered a negligible relationship as tendon stiffness could only account for 32% of the variability in impulse potentiation.



**Figure 15:** The scatterplots of maximal (tendon) stiffness and the various measures of potentiation in bench throws. A – Throw height potentiation, B – Impulse potentiation, C – EMG potentiation, D – F: EMG potentiation, E – Average power potentiation, F – Peak power potentiation ( $n = 19$ ).

All the submaximal stiffness measures were also compared to all of the bench throw data. The analysis was performed using case wise deletion for missing data, but was terminated where the subject numbers went below 10. The analysis of the submaximal stiffness data was therefore limited to a maximum load of 58.8 kg. No correlations were found between submaximal musculotendinous stiffness measures and any of the bench throw data for the 21.3 kg ( $n = 19$ ), 28.8 kg ( $n = 19$ ), 36.3 kg ( $n = 19$ ) and 58.8 kg ( $n = 11$ ) loads.

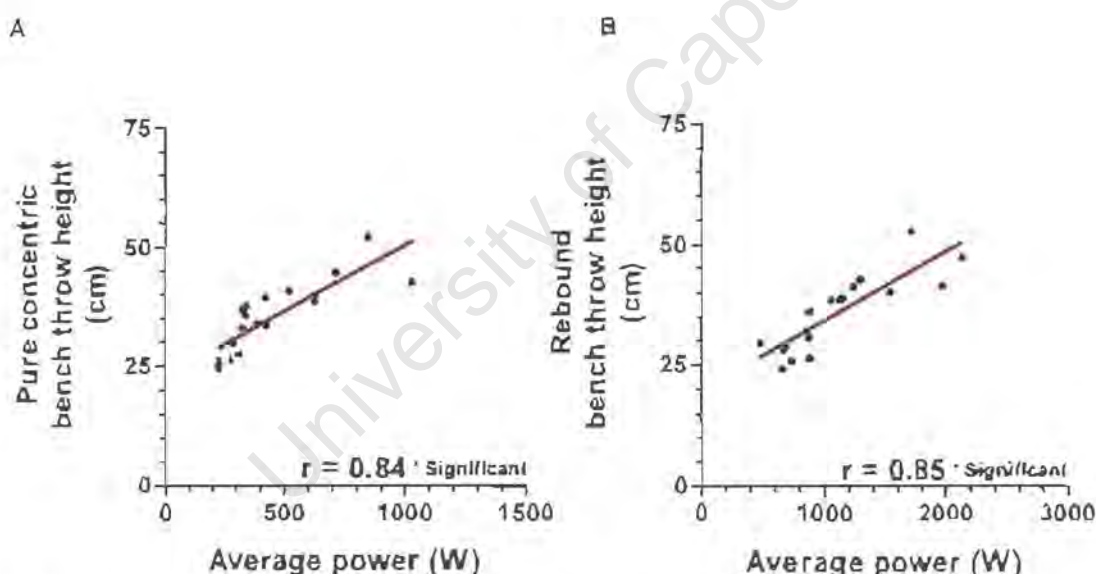
For the 43.8 kg load, the submaximal stiffness values were significantly correlated to pure concentric bench throw height ( $r = 0.48$ ,  $p < 0.04$ ) and average power of the rebound bench throws ( $r = 0.46$ ,  $p < 0.05$ ). For the 51.3 kg load, the submaximal stiffness values were significantly correlated to pure concentric bench throw height ( $r = 0.67$ ,  $p < 0.002$ ), rebound bench throw height ( $r = 0.63$ ,  $p < 0.004$ ), pure concentric bench throw impulse ( $r = 0.65$ ,  $p < 0.002$ ), rebound bench throw impulse ( $r = 0.62$ ,  $p < 0.005$ ), average power for pure concentric bench throws ( $r = 0.65$ ,  $p < 0.002$ ), average power for rebound bench



throws ( $r = 0.64$ ,  $p < 0.003$ ), peak power for pure concentric bench throws ( $r = 0.64$ ,  $p < 0.003$ ) and peak power for rebound throws ( $r = 0.62$ ,  $p < 0.004$ ). Even though musculotendinous stiffness at this load was significantly related to these abovementioned variables, the relationships were relatively weak.

### **BENCH THROW HEIGHT AND PERFORMANCE DATA**

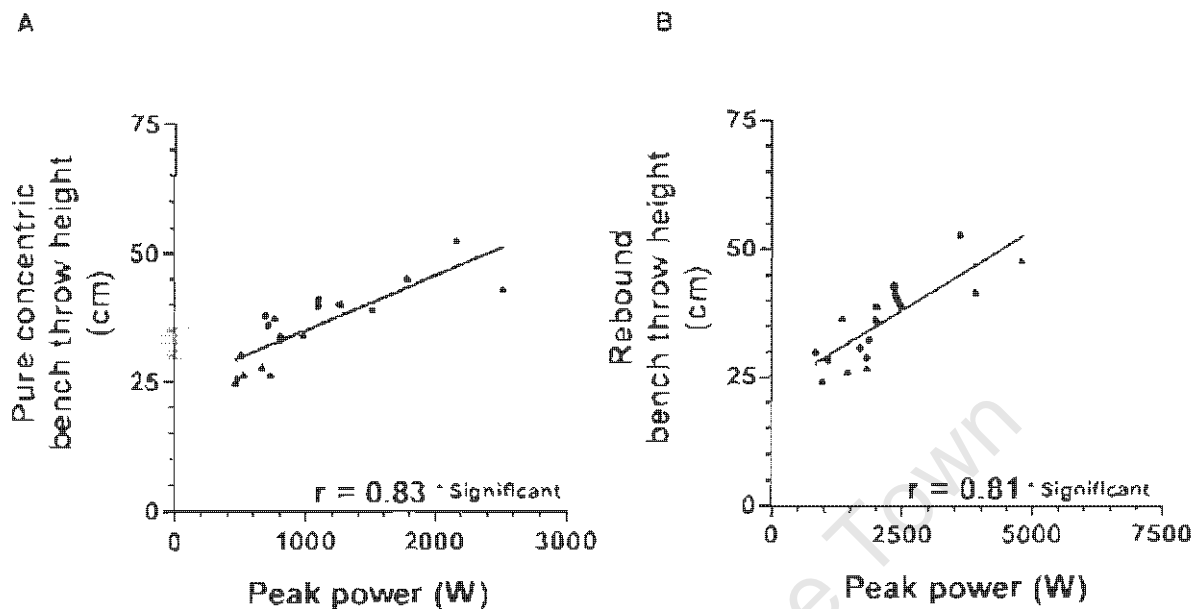
Average- (Figure 16) and peak power output data (Figure 17) using the modified regional breakdown and formulae of Harman et al. (69), with the calculated bench throw heights of both rebound and pure concentric throws, were compared. Figure 16 shows the scatterplot of pure concentric throw height (A) and rebound bench throw height (B) and their respective average power output data. Both (A) pure concentric throws ( $r = 0.84$ ,  $p < 0.0001$ ) and (B) rebound throws ( $r = 0.85$ ,  $p < 0.0001$ ) were correlated to the average power output of their respective force-time data.



**Figure 16:** The scatterplots of bench throw height and average power in pure concentric bench throws (A) and rebound bench throws (B) ( $n = 19$ ).

Figure 17 shows the scatterplots of pure concentric bench throw height (A) and rebound bench throw height (B) and their respective peak power output data. Both (A) pure concentric throws ( $r = 0.83$ ,  $p < 0.0001$ ) and (B) rebound throws ( $r = 0.81$ ,  $p < 0.0001$ ) were correlated to the peak power output of their respective force-time data. These abovementioned relationships were also reflected in their respective impulse data, which also correlated highly with pure concentric bench throw height ( $r = 0.85$ ,  $p < 0.0001$ ) and

rebound bench throw height ( $r = 0.87$ ,  $p < 0.0001$ ). Similar relationships were encountered with the average force data; pure concentric throws ( $r = 0.82$ ,  $p < 0.0001$ ) and rebound throws ( $r = 0.82$ ,  $p < 0.0001$ ).



**Figure 17:** The scatterplots of bench throw height and peak power in pure concentric bench throws (A) and rebound bench throws (B) ( $n = 19$ ).

## DISCUSSION

### MUSCULOTENDINOUS STIFFNESS AND STRETCH SHORTENING CYCLE POTENTIATION

The primary aim of this study was to explore the relationship between maximal musculotendinous (tendon) stiffness, as determined by the modified oscillation technique in the upper body, and stretch shortening cycle ability established by bench throws.

The maximal (tendon) stiffness data in the study ranged from  $8.3 - 35.2 \text{ kN.m}^{-1}$  averaging  $15.6 \pm 7.3 \text{ kN.m}^{-1}$ , which was similar to the values reported in previous research using a similar methodology (Chapter 3, p 160 - 161). With the modified upper body protocol, subjects managed on average between 5 - 6 loads and displayed a wide range of musculotendinous stiffness values. The damped oscillation frequency data also followed a



similar pattern to that demonstrated in our original studies, and is also consistent with the data available in the literature (Chapter 3, p 147). The oscillation frequency data ranged from 2.19 – 2.43 Hz over the various loads. The skill required to maintain static equilibrium and control during the oscillation test was lacking in 16% of the subjects initially recruited and screened for this study. This was a similar finding to the lower body study (Chapter 6). Irrespective of the strength of the subjects, it became progressively more difficult to maintain static equilibrium when the loads became heavier and frequently fewer than three force oscillations were obtained for the calculation of stiffness. However, even under these conditions, the same stringent controls previously utilised ensured that reliable data were acquired and analysed.

Previous studies comparing maximal musculotendinous (tendon) stiffness and stretch shortening cycle improvement of muscle performance in the upper body have used impulse (defined by area under the force-time curve) (156), and loads lifted to quantify the stretch shortening cycle improvement of performance (42;153;156). This study adopted a similar protocol to that used in the lower body study. Unlike the other studies comparing musculotendinous stiffness in the upper body (42;153;156), this study used the same load i.e. the combined mass of the bar and linear bearing tracking-unit (21.32 kg) in both bench press variations. This study utilised pure concentric- and rebound bench throws in a similar procedure to that adopted by Newton et al. (119). The authors had previously shown the bench throws to be analogous to the “vertical jumps” of the upper body (Chapter 1, p 37 - 38).

This study showed no appreciable relationships between tendon stiffness and all but one of the stretch shortening cycle potentiation measures (Table 5). Consistent with previous findings (156), we found a significant inverse ( $r = -0.57$ ,  $p < 0.01$ ) relationship between tendon stiffness and impulse potentiation (Figure 15). This was, however, a weak association and tendon stiffness could only account for 32% of the variation in muscle performance gain. The relationship shown in this study was also considerably lower than previously reported (156). Wilson et al. (156) observed a significant inverse relationship ( $r = -0.72$ ,  $p < 0.01$ ) in the upper body between maximal or tendon stiffness and the augmentation in concentric impulse obtained during the stretch shortening cycle or rebound bench press.

As with the lower body, one could not accurately differentiate between eccentric and concentric phases of stretch shortening cycle muscle action in the bench throw data (Chapter 5). Therefore the muscle performance data were representative of the combined eccentric and concentric phases of contraction. However, because the eccentric contraction in the stretch shortening cycle allows the muscle-tendon complex to initiate concentric contraction from a higher force level (18), we have to assume that the combined force data also mirrors the concentric impulse potentiation sufficiently to enable comparison.

In addition, this study used lighter loads than the aforementioned study of Wilson et al. (156). Wilson et al. (156) showed potentiation in concentric impulse over the first 370 ms of concentric contraction. It is feasible that because of the heavy loads moved in their study, the speed of concentric contraction was reduced. This, in turn, could have confined the effects of stretch shortening cycle potentiation to the initial portion of the concentric phase purely due to the extended duration of the movement. In this study, we utilised a lighter load, which would increase the speed and decrease the duration of the concentric movement. It is therefore reasonable to assume that the potentiation in concentric impulse might have extended over a greater proportion of the concentric propulsion phase and that the combined data would therefore reflect this.

Furthermore, Wilson et al. (156) never used the same loads to compare their data. For their pure concentric bench press and rebound bench press tests, they used 95% of their respective 1RM test loads - the rebound bench press being significantly greater. Doan et al. (42) showed that by increasing the eccentric load in the upper body bench press movement, one could significantly improve the ensuing concentric performance i.e. the weight lifted. As a result of the increased weight that Wilson et al. (156) utilised in the rebound bench press- as opposed to the pure concentric bench press test, and simply due to the fact that different loads were used for the respective tests, it may be that the concentric augmentation in impulse via the stretch shortening cycle might have been exaggerated. This in turn could have resulted in an over-estimation of the increase in impulse and possibly also the relationship between maximal musculotendinous (tendon) stiffness and impulse potentiation. Nonetheless, tendon stiffness in their study could only explain 51% of the variance in stretch shortening cycle improvement, which is not a strong relationship.

The submaximal musculotendinous stiffness values were also compared to the stretch shortening cycle measures. No relationships were found between any of the submaximal musculotendinous stiffness- and stretch shortening cycle potentiation measures.

The musculotendinous stiffness data in the upper body, using a multi-joint testing procedure similar to that of Wilson et al. (153;155;156), confirm a weak relationship between maximal stiffness and impulse potentiation and show no relationship to any of the other measures of stretch shortening cycle potentiation (Table 5). As with the lower body study (Chapter 6), we therefore question the ability of the isolated measurement of tendon stiffness in explaining stretch shortening cycle potentiation of performance in the upper body.

### **MUSCULOTENDINOUS STIFFNESS AND BENCH THROW PERFORMANCE**

This study showed significant relationships between maximal musculotendinous (tendon) stiffness and both pure concentric- and rebound bench throw height. These relationships accounted for 45 and 51% of the variability in pure concentric- and rebound bench throw height respectively. This suggests that a stiffer tendon might partially contribute towards a better bench throw height, regardless of the muscle action. This could in part be due to the training status of the subjects, as it has been stated that stronger individuals have stiffer and stronger tendons (115;153). Bench throw height could to a very limited degree therefore be an expression of the strength of the subjects.

Similar relationships were shown between maximal (tendon) stiffness and both average- and peak power outputs of both pure concentric bench throws and rebound bench throws. Tendon stiffness could describe 69% and 50% of the variability in average power in pure concentric throws and rebound throws respectively (Figure 9). There was a similar trend with peak power where tendon stiffness could explain 76% and 62% of the variability in muscle performance in that order (Figure 10). There seems to be a noticeable relationship between tendon stiffness and the power developed in especially the pure concentric bench throws. The relationship between tendon stiffness and power in rebound bench throws was not as pronounced. The abovementioned results imply that a stiffer tendon allows for a greater average- and peak power output, especially in pure concentric bench throws.

Considering the relationship between the rate of force development and power output, these results seem to be in accordance with previous research findings (151;152;155).

Subsequently the submaximal stiffness data at all levels (where the subject numbers did not decrease below 10) were compared with bench throw heights, average power outputs, peak power outputs, and all the other kinetic variables in both pure concentric- and rebound bench throw conditions. The submaximal stiffness data at the lesser loads were not related to any of these variables. However weak, but significant correlations were found at the 43.8 kg load between musculotendinous stiffness and pure concentric throw height and average power of the rebound bench throws. There were also associations between submaximal musculotendinous stiffness at the 51.3 kg load and bench throw height, impulse, average power and peak power in both conditions. The associations were however very weak as stiffness could only explain between 38 – 44% of the variability in the aforementioned measures. The 58.8 kg load, which followed, indicated no noticeable relationships between musculotendinous stiffness and any of the bench-throw data. This might however have been an artefact of the reduced number of subjects in this data set.

### **REBOUND BENCH THROWS VS PURE CONCENTRIC BENCH THROWS**

The bench throw data followed a similar pattern to original research using a comparable procedure to quantify stretch shortening cycle potentiation (119). In this study there was no potentiation in functional performance data, i.e. bench throw height in rebound throws as opposed to pure concentric throws (Figure 11). There was substantial enhancement in average- and peak power (Figure 12), and average force by means of the stretch shortening cycle. These findings are also in accordance with those of Newton et al. (119). Furthermore, in agreement with the lower body data (Chapter 6), the stretch shortening cycle affected power more than it affected force.

Newton et al. (119) compared bench throws at various percentages of maximum (1RM) rebound bench press i.e. 15, 30, 45, 60, 75 and 90%. Even though they found significant increases in average velocity, average- and peak force, average power and peak power in stretch shortening cycle throws, this did not convert into a functional improvement in performance i.e. a potentiation in throw height. The height thrown during rebound throws

and pure concentric throws over their range of loads, as with this study, did not significantly differ from one another. Newton et al. (119) attributed this lack of functional improvement to the fact that recovery of elastic energy and utilisation of stretch reflex facilitation only contributed to the initial phases of the throw, and due to the relatively increased range of motion and longer duration of the throws, its effect was diminished during the latter stages of the movement.

This study utilised a relatively light load (21.32 kg), therefore the duration of the throw would not necessarily have had as much influence. The lack of improvement in functional performance might also have been related to the large range of motion utilised (29;119;141). Because the depth of the rebound or lowering movement was restricted to just above the chest for comparative purposes (Chapter 2, p 127), subjects could not self-select their natural or optimal range of movement. This might also have adversely affected functional performance in the rebound bench throws.

Despite the lack of improvement in functional performance with rebound bench throws versus pure concentric bench throws, significant increases occurred in impulse and EMG (Figure 13), and the F: EMG ratio (represented by dividing the integrated force/impulse data by the integrated EMG data) (Figure 14); impulse increased by  $64 \pm 12\%$ , EMG by  $11 \pm 15\%$  and F: EMG ratio by  $49 \pm 25\%$ . In the upper body it was apparent that even though functional performance i.e. throw height did not improve, muscle performance i.e. force and power output did. The improved force and power output could not however be explained via increased muscle activation (EMG) alone. The F: EMG ratio improved considerably, despite significant individual increases in both of its subcomponents i.e. EMG and impulse. On the basis of these findings, as with the lower body (Chapter 6), it would therefore seem that there was a significant elastic contribution to muscle performance enhancement (23).

Given the methodology utilised in this study, it may be concluded that the elastic nature of the muscle-tendon complex in the upper body acts as a power amplifier rather than as a mechanism to improve functional performance i.e. heights thrown. One therefore has to concur with Newton et al. (119) that the range of motion might be too large to effectively transfer the positive stretch shortening cycle effects into increased throw heights.

In support of Newton et al. (119), it further seems likely that the effects of the stretch shortening cycle contribute substantially more to the initial portion of the concentric movement or propulsion phase. In this way it would assist in raising the bar beyond the 'sticking point' and thereby contribute to increasing functional performance rather by increasing the load lifted than by increasing the height that the bar is thrown. This interpretation seems to be supported by the findings in this study and also by previous research (42;119;153;156). One cannot, however, disregard the possible contributions to this phenomenon from other sources e.g. the interaction effects between muscle and tendon, and the force-acceleration effect (discussed in detail in Chapter 1), which were not measured in this study.

### ***BENCH THROW HEIGHT AND PERFORMANCE DATA***

As with the lower body data (Chapter 6), the force-plate data correlated well with the functional performance data of both pure concentric bench throws and rebound bench throws. Pure concentric throw height and rebound throw height were significantly correlated to their respective average power output, peak power output, net impulse and average force data measured on the custom-built force plate. These measures could account for 66-76% of the variance in bench throw heights.

It therefore seems that the upper body bench throws measured with the NAMS unit are consistent with, the results of the lower body data. Therefore, on the basis of the abovementioned relationships, it may be concluded that the force-plate data are an accurate representation of muscle function in the upper body.

### **CONCLUSION**

As with the previous study in the lower body (Chapter 6), an attempt was made to isolate tendon elasticity (stiffness) in the upper body. Tendon elasticity was next related to the associated stretch shortening cycle performance of the upper body musculature. It may be concluded that tendon elasticity, as an isolated component of muscle function, is poorly related to stretch shortening cycle performance in the upper body. On the basis of the almost negligible relationship between tendon elasticity and impulse potentiation, it can be argued that the isolated measurement of tendon elasticity has little value in explaining

stretch shortening cycle performance and understanding the mechanisms involved. As with the lower body, further research in this area should focus on the integration of, and combined interaction between, the numerous mechanisms involved in performing stretch shortening cycle activities. This should provide a better basis for understanding dynamic muscle function and individual stretch shortening cycle enhancement of performance.

University of Cape Town

## CHAPTER 8:

### Final summary and conclusions

#### RESEARCH QUESTION

The primary goal of this dissertation was to identify the relationship between the mechanical characteristics of the muscle-tendon complex, in particular tendon stiffness, and stretch shortening cycle muscle function. To fulfil this goal, the main aims of this thesis were:

- (i) To design and develop a testing procedure/equipment that could measure stretch shortening cycle muscle function and musculotendinous stiffness in both upper and lower body musculature
- (ii) To determine if the procedures for measuring musculotendinous stiffness in both upper and lower body musculature were repeatable
- (iii) To determine if the procedures for measuring stretch shortening cycle muscle function in both upper and lower body musculature were repeatable
- (iv) To refine the testing procedures, based on the experience gained in (ii) and (iii)
- (v) To determine the relationship between musculotendinous stiffness and stretch shortening cycle muscle function in both upper and lower body musculature

The following section attempts to provide a brief, summarised answer to each research question presented above.

- (i) *To design and develop a testing procedure/equipment that can measure stretch shortening cycle muscle function and musculotendinous stiffness in both upper and lower body musculature*

The NAMS unit (**N**euromuscular **A**nd **M**usculotendinous **S**tiffness unit) was designed and built in collaboration with Zest Manufacturing PTY (Ltd), Cape Town, South Africa, to measure stretch shortening cycle muscle function and musculotendinous stiffness in both upper and lower body musculature. The NAMS unit is a custom-built computerised system consisting of a modified Smith-machine, a force-plate, and an attachable leg press unit with a load cell, a leg press seat section, and an adjustable bench press bench.



The NAMS unit was completed and subjected to procedural and pilot testing in July 2001.

- (ii) *To determine if the procedures for measuring musculotendinous stiffness in both upper and lower body musculature are repeatable*

The oscillation technique, which was used to measure musculotendinous stiffness in the upper and lower body extremities with the NAMS unit, was repeatable. Musculotendinous stiffness in the lower body musculature differentiates strong from weak individuals rather than relatively stiff from relatively compliant individuals. To overcome this it is recommended that musculotendinous stiffness be corrected for load and body mass for relative comparisons between subjects.

In the upper body musculature there was a lesser relationship between load and stiffness compared to the lower body.

In summary, the oscillation technique was shown to be a reliable measure of musculotendinous stiffness in both lower and upper body tests. However, the determination of musculotendinous stiffness in this format was influenced by the load against which the muscle was contracting, and the subject's body mass in the muscles of the lower body, and to a lesser extent in the muscles of the upper body. The technique needed to be refined to address these, and some additional procedural concerns.

- (iii) *To determine if the procedures for measuring stretch shortening cycle muscle function in both upper and lower body musculature are repeatable*

The first main conclusion of the lower body study was that the measurements of stretch shortening cycle performance of the lower body were generally repeatable when measured with the NAMS unit. However, the "Concentric-defined" measurements were less repeatable and should be used with caution.

The second main conclusion of this study was that even though many of the actual data or individual measurements of stretch shortening cycle performance were repeatable as

individual entities, when they were incorporated into ratios and mathematical derivatives, the data were no longer necessarily repeatable or reliable.

The main conclusion of the upper body study was that the measurements of stretch shortening cycle performance of the upper body were generally repeatable when measured with the NAMS unit. However, as with the lower body study, the "*Concentric-defined*" measurements, especially in the upper body, lacked sufficient repeatability and had a higher variability and therefore should also be used with caution.

As with the lower body data, many of the individual measurements of stretch shortening cycle performance were repeatable. However, when these measurements were incorporated into ratios and mathematical derivatives, the data were also no longer necessarily repeatable or reliable.

In summary, the vertical jump height and bench throw height measures using the NAMS unit were repeatable and had sufficient precision to detect changes with about 10 subjects. However, when familiarizing subjects with these techniques more attention needs to be applied to the stretch shortening cycle tests i.e. countermovement jumps and rebound bench throws. Additionally with the rebound bench throws, more rest between repetitions and tests needs to be given to prevent fatigue from influencing the rebound throw height and thereby improving the accuracy of the measure. It might also be prudent to use this same approach in the lower body test.

The MVC peak force, MVC peak load and MVC maximal effort measures i.e. MVC impulse and MVC EMG, of both the upper and lower body tests using the NAMS unit were all also repeatable.

The "Total" force and "Total" IEMG measures of both upper and lower body tests were also repeatable. However, the "*Concentric-defined*" force and IEMG measures of both upper and lower body measures were not as repeatable and should not be used with a relatively small sample size ( $\pm 10$  subjects).

One should also be cautious in the interpretation of stretch shortening cycle potentiation measures. The mathematical manipulation of very repeatable data sets e.g. squat jump

height and countermovement jump height in the lower body, and pure concentric bench throw height and rebound bench throw height in the upper body, seems to be flawed when combined and expressed as differences, ratios or relative change indices, as this reduces the reliability and increases the variability of these derived measures. To represent the potentiation in muscle performance as a result of the stretch shortening cycle muscle action, one however needs to relate the pure concentric measures with the stretch shortening cycle measures, and at present there does not seem to be a better alternative in quantifying this relationship. One should however, in future research, acknowledge and understand the limitations of the derived measures of potentiation expressed in this way.

- (iv) *To refine the testing procedures, based on the experience gained in (ii) and (iii)*

Certain methodological shortcomings in the original oscillation technique utilised in the repeatability studies were identified. Although the measure of stiffness of the muscle-tendon complex using the NAMS unit was repeatable, these methodological concerns needed attention, before the technique could be used in applied or mechanistic experiments.

The oscillation technique was modified to use an absolute rather than a relative loading protocol, and the formulation for deriving maximal stiffness was changed from using the downward exponential association equation, to the Boltzmann sigmoid equation. This revised method seemed to be a more accurate method of measuring the stiffness-load relationship and musculotendinous stiffness, than the original method.

The maximum voluntary contraction procedure used in the original technique underestimated the maximum strength in some subjects. Using absolute loads and increasing the load until the subject could not maintain it with sufficient stability seemed to have accounted for this error. It was therefore proposed that further studies using the oscillation technique should follow an absolute loading rather than a relative loading protocol.

In the repeatability studies on stretch shortening cycle measures in both lower and upper body studies, it was evident that the measures used to describe potentiation after stretch

shortening cycle activity e.g. percentage jump/throw height improvement, as they have been described in previous studies, lacked repeatability, even though their individual measures e.g. squat jump/pure concentric bench throw height and countermovement jump/rebound bench throw height, were repeatable. This raised the question of the most appropriate way to accurately measure stretch shortening cycle potentiation.

In an attempt to answer this question, the testing procedure was analysed, given the experience gained in the original repeatability studies. Based on logical reasoning, the following changes were made to the testing protocol: In the initial testing procedure, there was no normalisation for either the additional body- and bar weight in the lower body study, or for the additional bar weight in the upper body study. The data gathered were nevertheless repeatable. However, human judgment was often involved in determining the initiation of contraction, the take-off/release and the landing/catch on the force plate, where vibration from the top plate/body hindered the accuracy of the measurement in the lower body/upper body trials respectively. It is logical to assume that by eliminating experimental error the repeatability of these measures would improve resulting in greater accuracy of the measures of potentiation or mathematical derivatives.

Therefore the protocols were adapted by zeroing the force-measurement of the force-plate to accommodate the additional weight of the subject and bar in the lower body studies, and the additional weight of the bar in the upper body studies to follow, thereby normalising each individual for the additional subject-bar or bar weight respectively. Additionally the force-time data segments of muscle function were adjusted to accommodate these changes.

- (v) *To determine the relationship between musculotendinous stiffness and stretch shortening cycle muscle function in both upper and lower body musculature*

An attempt was made in the first study to isolate tendon elasticity (stiffness) in the lower body. Tendon elasticity is one of the key components of muscle function. Next, tendon elasticity was related to the associated stretch shortening cycle performance.

It was concluded that tendon elasticity, as an isolated component of muscle function, was not related to the associated stretch shortening cycle muscle performance in the lower body.

Another interpretation of these data is that there are numerous factors, which contribute to muscle function that are interlinked and involved in the enhancement of performance observed in stretch shortening cycle movements. Further research focusing on stretch shortening cycle mechanisms in the lower body should move towards an integrated analysis approach examining the combined effect of multiple systems and their interaction with each other during dynamic muscle function.

As with the lower body, an attempt was made to isolate tendon elasticity (stiffness) in the upper body. Tendon elasticity was next related to the associated stretch shortening cycle performance of the upper body musculature.

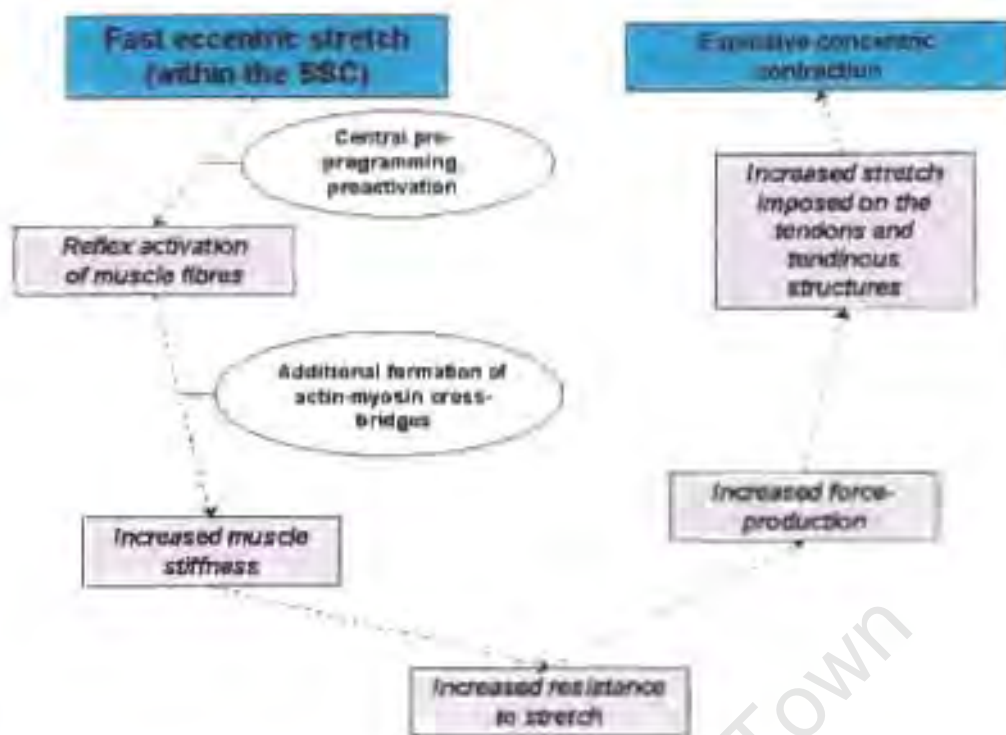
It was concluded that tendon elasticity, as an isolated component of muscle function, had a weak relationship to stretch shortening cycle performance in the upper body. On the basis of the weak relationship between tendon elasticity and impulse potentiation in the upper body, it can be argued that the isolated measurement of tendon elasticity has little value in explaining stretch shortening cycle performance and the mechanisms involved. As with the lower body, further research in this area should focus on the integration of, and combined interaction between, the numerous mechanisms involved. This should provide a better basis for understanding dynamic muscle function and individual stretch shortening cycle enhancement of performance.

# **OVERALL CONCLUSIONS OF THESIS**

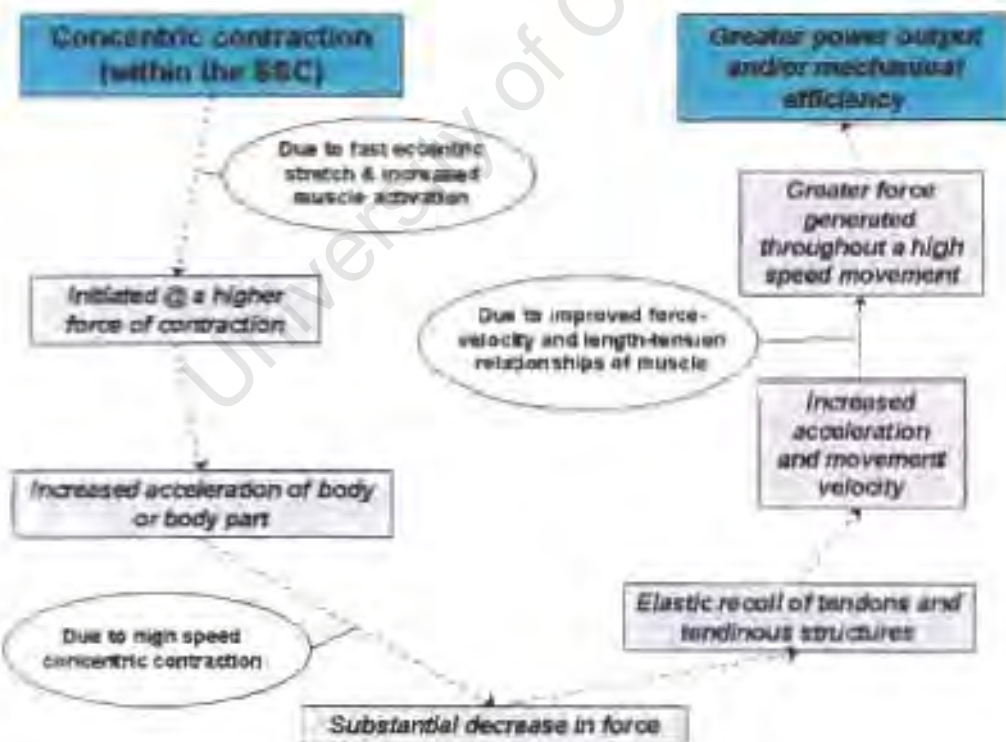
## ***THE STRETCH SHORTENING CYCLE PROCESS***

On the basis of the literature review (Chapter 1), the following assumptions were made with regards to the stretch shortening cycle enhancement of performance:

- During eccentric stretching of muscle (Figure 1), the reflex pathways are stimulated, which as a protective mechanism, increases neural activation and the formation of additional actin-myosin cross-bridges.
- This facilitates an increase in muscle stiffness and the resistance to stretch in the involved muscle fibres.
- This increased muscle stiffness and resistance to stretch boosts the force generated in the muscles, and imposes the majority of the stretch on their respective tendons and aponeuroses.
- Concentric contraction (Figure 2) is initiated at a higher force and activation level, which accelerates the body or body part faster and more forcefully, than it would during isolated concentric contraction.
- As a result of the substantial decrease in force from eccentric to concentric contractions, combined with the increased acceleration during the initial part of concentric contraction, the recoil of the previously stretched tendons and aponeuroses, additionally assists in the shortening of the muscle-tendon complex.
- Furthermore, because of the complex interaction between muscle and tendon, the muscle fibres, due to more optimal length-tension and force-velocity relationships, also contract more forcibly throughout the movement.
- The combination of events rather than the isolation of any of these abovementioned mechanisms contribute to enhance the power output and mechanical efficiency found in stretch shortening cycle movements.



**Figure 1:** The stepwise progression of the eccentric contraction phase of the stretch shortening cycle muscle action.



**Figure 2:** The stepwise progression of the concentric contraction phase during the stretch shortening cycle muscle action.

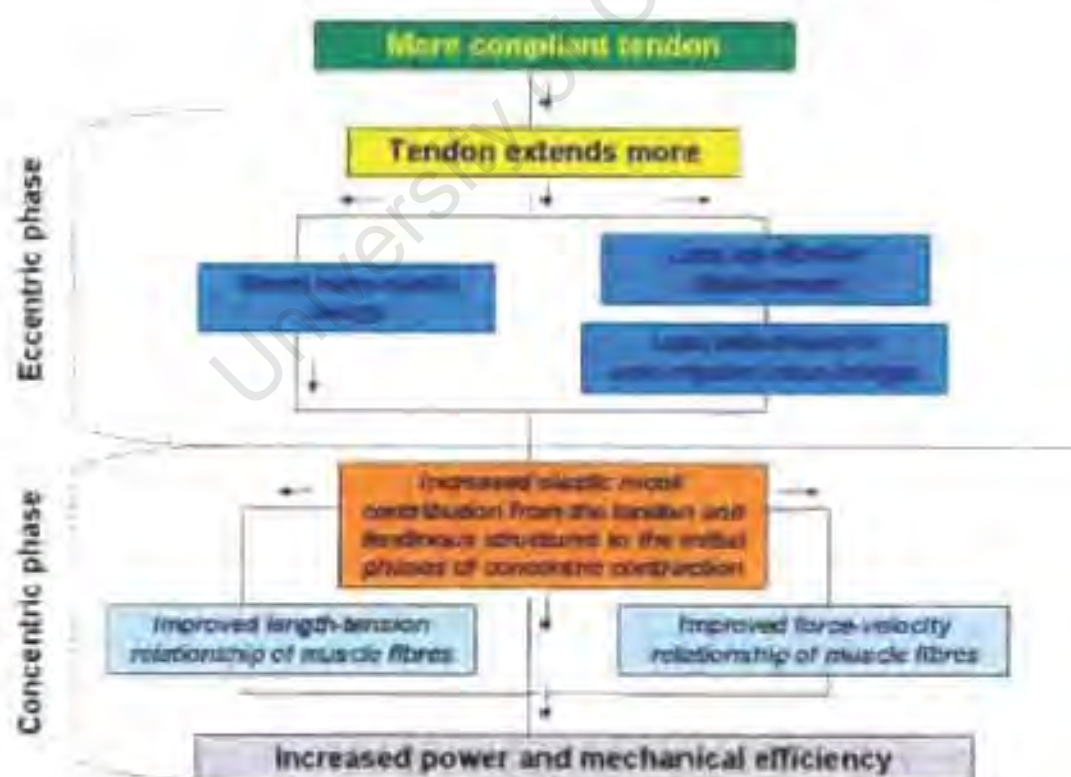


## TENDON ELASTICITY AND STRETCH SHORTENING CYCLE PERFORMANCE

Chapters 6 and 7 show that there is no, or a weak relationship between tendon elasticity (stiffness) and stretch shortening cycle potentiation of performance in the lower and upper body respectively. In the literature review (Chapter 1) it was suggested that tendon stiffness could affect stretch shortening cycle potentiation of muscle performance.

### A MORE COMPLIANT TENDON

It is possible that a muscle-tendon complex with a more compliant tendon (Figure 3) elicits stretch shortening cycle potentiation of muscle performance primarily as a result of the increased elastic recoil assistance from the tendons and tendinous structures, combined with the enhanced interaction effects associated with a more compliant tendon, i.e. improved force-velocity and length-tension relationships of the involved muscle fibres.

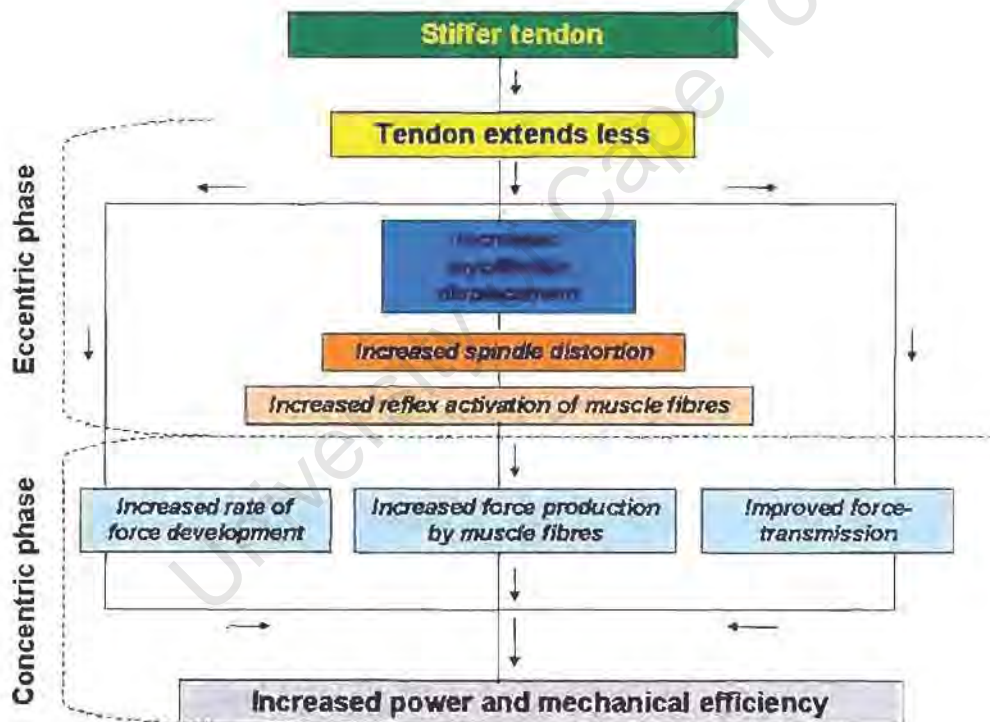


**Figure 3:** The proposed mechanisms through which a muscle-tendon complex with a more compliant tendon elicits improved muscle performance via the stretch shortening cycle.



## A STIFFER TENDON

Alternatively, it is plausible that a muscle-tendon complex with a stiffer tendon (Figure 4) elicits stretch shortening cycle potentiation of muscle performance primarily via enhanced neural mechanisms and improved force-transmission capabilities. It is possible that the increased linear extension of the involved myofibrils during the eccentric contraction phase, and the corresponding increased muscle spindle distortion associated with a stiffer tendon, enhances the reflex activation of additional muscle fibres, and/or the formation of additional actomyosin cross-bridges within the already activated muscle fibres. This enhanced muscle activation coupled with the better force-transmission capabilities of a stiffer muscle-tendon complex, could enhance the concentric force and power output of the muscle-tendon complex during the stretch shortening cycle muscle action.



**Figure 4:** The proposed mechanisms through which a muscle-tendon complex with a stiffer tendon elicits improved muscle performance via the stretch shortening cycle

## FINAL CONCLUSIONS

In the lower body (Chapter 6) and upper body (Chapter 7) studies, no moderate to strong relationships were shown between maximal musculotendinous (tendon) stiffness and stretch shortening cycle potentiation of any of the involved measurements i.e. jump/throw

height, average- and peak power, impulse, EMG and F: EMG ratios. A small, almost negligible inverse relationship was found between tendon stiffness and impulse potentiation in the upper body study (Chapter 7). A limitation of the technology and methodology used in these studies was that (i) it lacked sufficient sensitivity to accurately and reliably differentiate between eccentric and concentric phases, (ii) it was unable to directly measure internal musculotendinous processes, and (iii) it could not measure central preactivation and reflex involvement. Therefore, it is difficult to quantify their specific contributions to the stretch shortening cycle potentiation of muscle performance.

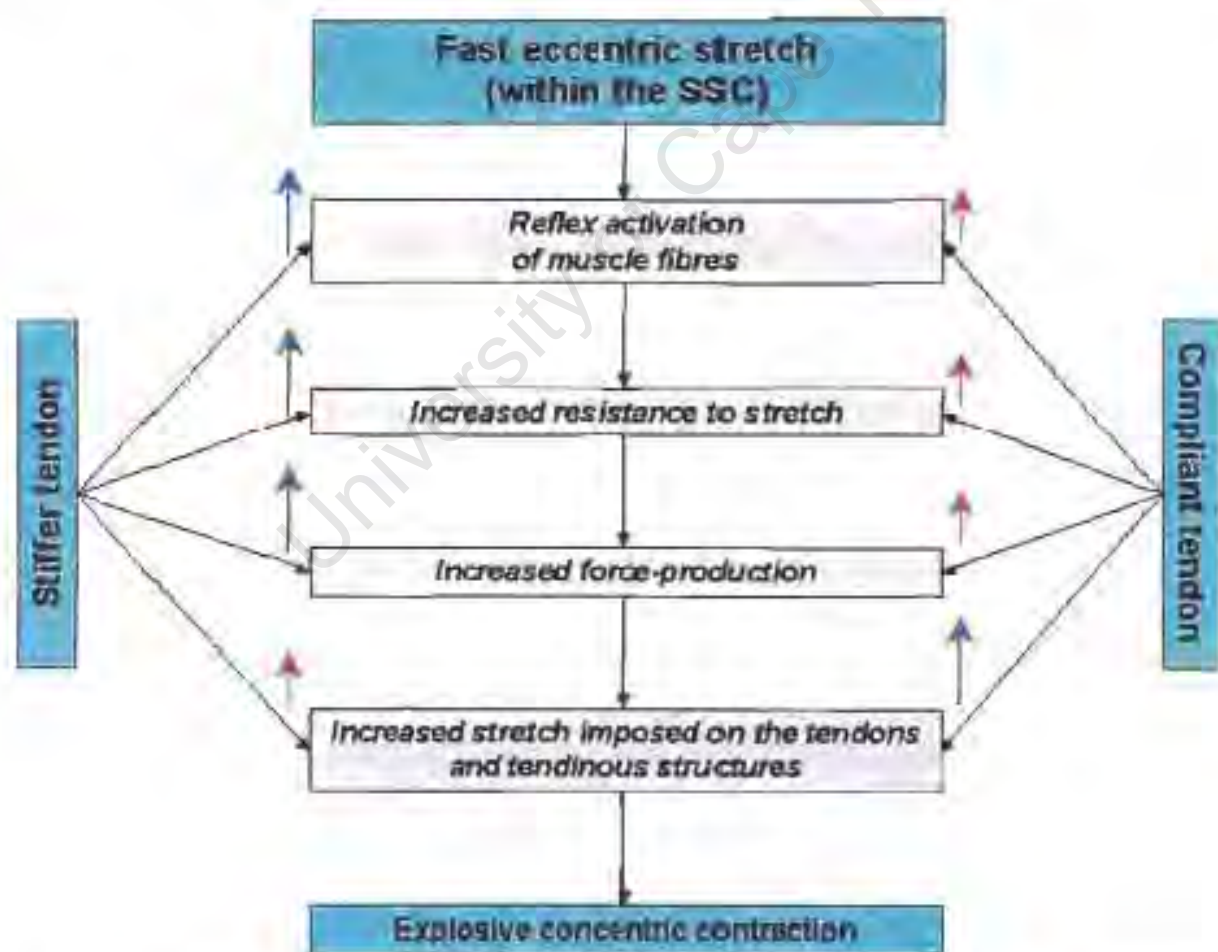
Nevertheless, numerous muscle performance measures showed significant potentiation during the stretch shortening cycle muscle action, and these were not related to tendon stiffness. It therefore seems that the stretch shortening cycle potentiation of muscle performance is a protected variable in dynamic muscle function. It further seems that this potentiation of muscle performance occurs regardless of different muscle and tendon elastic characteristics. Accordingly it seems logical to assume that different combinations of varying proportional neuromuscular and elastic contributions (described in Figures 3 and 4) could elicit equal stretch shortening cycle potentiation of muscle performance.

### **HYPOTHESES RELATING TO THE STRETCH SHORTENING CYCLE MUSCLE ACTION**

Figures 5 and 6 illustrate the theoretical opposite ends of the tendon elasticity continuum, both leading to equal stretch shortening cycle potentiation of muscle performance. The following are theoretical hypotheses on the understanding of the effects of different tendon stiffness extremes and their respective interactions to elicit equal stretch shortening cycle potentiation of muscle performance (Figures 5 and 6). It may be proposed that in both extremes i.e. a very stiff and a very compliant tendon that all the stepwise interactions illustrated in Figures 1 and 2 occur during the stretch shortening cycle muscle action regardless of their different tendon elastic characteristics. However, it may be speculated that the proportional contributions of these mechanisms to the stretch shortening cycle potentiation of muscle performance will vary according to their respective tendon characteristics.

## THE ECCENTRIC CONTRACTION PHASE

Figure 5 describes the projected interactions within the eccentric contraction phase of the stretch shortening cycle muscle action. As a result of the mechanisms depicted in Figure 4, for a muscle-tendon complex with a stiffer tendon, a greater reflex activation response of the involved muscle fibres in resistance to stretch is expected compared to a muscle-tendon complex with a more compliant tendon. This would correspondingly lead to greater muscle stiffness and therefore greater resistance to stretch, thereby increasing the force production in the stiffer muscle-tendon complex to a greater extent than in a muscle-tendon complex with a more compliant tendon (Figure 5). Both the 'stiffer tendon' and 'more compliant tendon' complexes would impose a certain amount of stretch on their respective tendons, yet for the same amount of loading it may be proposed that the strain would be greater in the more compliant muscle-tendon complex (Figure 5).



**Figure 5:** A comparative theoretical projection of the effects of a stiffer vs. more compliant tendon on the eccentric contraction phase of stretch shortening cycle movements.

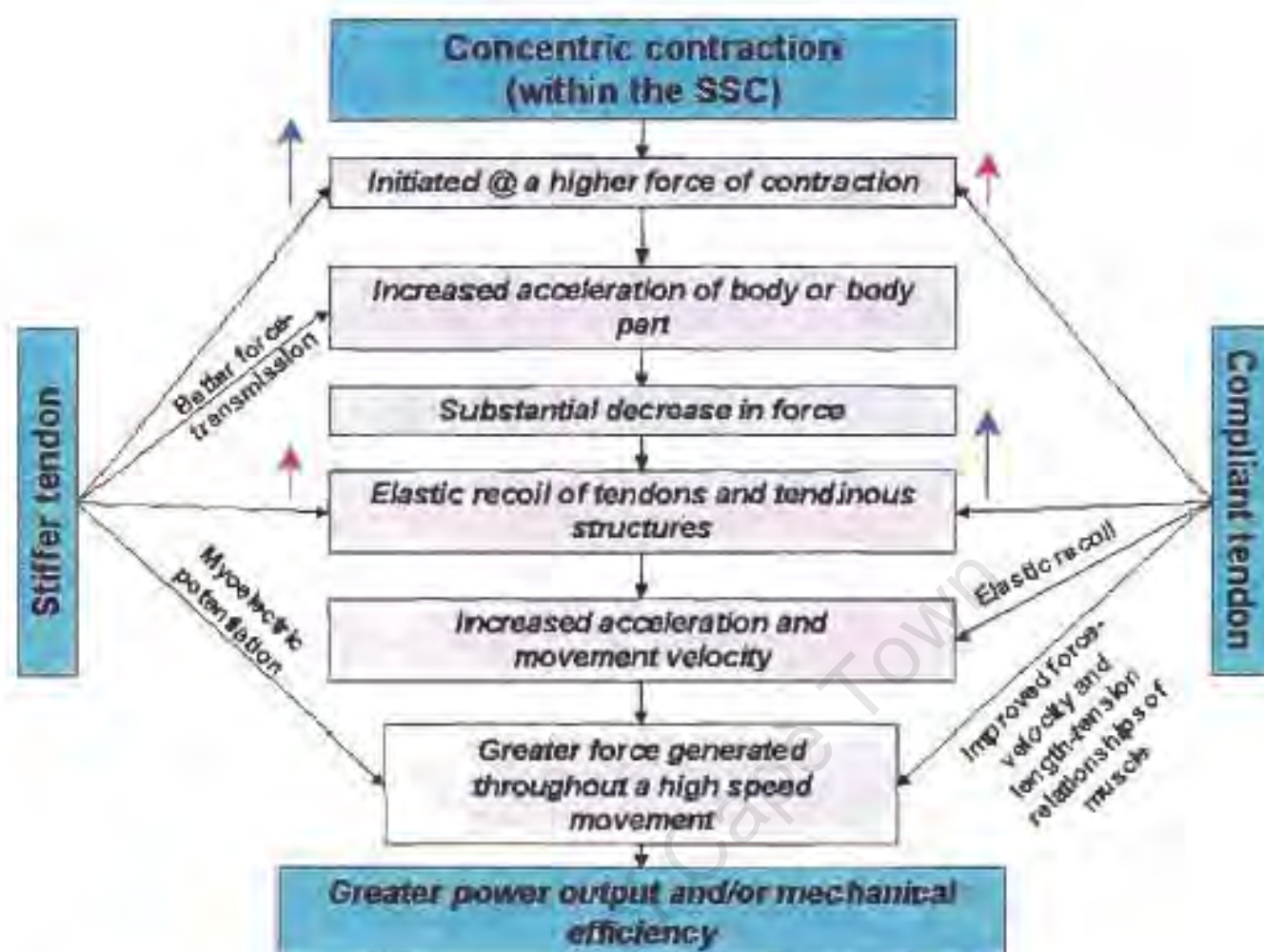


## THE CONCENTRIC CONTRACTION PHASE

Figure 6 describes the projected interactions within the concentric contraction phase of the stretch shortening cycle muscle action.

Due to the mechanisms described in Figure 5, it may be speculated that for a muscle-tendon complex with a stiffer tendon, concentric contraction would be initiated from a higher force level than for a muscle-tendon complex with a more compliant tendon (Figure 6). In a muscle-tendon complex with a stiffer tendon compared to a muscle-tendon complex with a more compliant tendon, this increased force at the start of the concentric contraction phase, coupled with the better force-transmission capabilities leads to an increased acceleration of the body or body part during the initial phase of the concentric contraction. There is still an elastic recoil contribution from the tendons and tendinous structures and muscle-tendon interaction, yet the contribution from these sources towards the stretch shortening cycle potentiation of muscle performance would be significantly less in a stiffer system than for a more compliant system (Figure 6). The greater muscle activation as a result of eccentric pre-stretch in a stiffer system compared to a more compliant system potentiates force-production throughout the remainder of the concentric phase of the stretch shortening cycle muscle action, thereby increasing power output and mechanical efficiency.

For a muscle-tendon complex with a more compliant tendon, and due to the mechanisms represented in Figure 3 and described in Figure 5, concentric contraction would also be initiated from a higher force level, albeit to a lesser extent than for a stiffer complex. However, as a result of the increased strain of the tendons and tendinous structures due to eccentric pre-stretch during the stretch shortening cycle muscle action (Figure 5), the tendons and tendinous structures respond accordingly by recoiling elastically at the onset of the concentric contraction to a greater extent than a muscle-tendon complex with a stiffer tendon. This accelerates the initial part of the concentric movement, which allows the muscle fibres to function at better length-tension and force-velocity relations than a muscle-tendon complex with a stiffer tendon, and thereby allows them to contract more forcefully. The combination of these mechanisms contributes to the improved power and mechanical efficiency during the stretch shortening cycle muscle action in a muscle-tendon complex with a more compliant tendon.



**Figure 5:** A comparative theoretical projection of the effects of a stiffer vs. more compliant tendon on the eccentric contraction phase of stretch shortening cycle movements.

It needs to be reiterated that the ideas in this section have evolved from the results of the various studies in this thesis and need to be examined systematically in future studies for verification.

## FINALE

The relative contributions of these various mechanisms to potentiate muscle performance during the stretch shortening cycle muscle action, seems to be more complex than originally expected. Further research needs to focus on the combined functioning and interaction between the mechanisms involved in the stretch shortening cycle potentiation of muscle performance to understand the complexity of this dynamic form of muscle function. The combination of events rather than any one specific mechanism is ultimately

responsible for the improved power and mechanical efficiency shown during the stretch shortening cycle muscle action.

University of Cape Town

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University of Cape Town

## **APPENDICES**

University of Cape Town

UNIVERSITY OF CAPE TOWN



*Research Ethics Committee*  
Faculty of Health Sciences  
Anzio Road, Observatory, 7925  
Queries : Xolile Fula  
Tel : (021) 406-6492 Fax: (021) 406-6390  
E-mail : Xfula@curie.uct.ac.za

02 July 2001

REC REF: 086/2001

Mr LW Viljoen  
Human Biology

Dear Mr Viljoen

**THE EFFECT OF TENDON STIFFNESS ON STRETCH SHORTENING CYCLE  
PERFORMANCE**

Thank you for your application submitted to the Research Ethics Committee on the 06 March 2001.

I have pleasure in informing you that the above study has been formally approved by the Research Ethics Committee on the 14 June 2001.

Please quote above REC reference number in all correspondence.

**Signed by candidate**

PROFESSOR CR SWANEPOEL  
CHAIRPERSON

**UNIVERSITY OF CAPE TOWN**



Research Ethics Committee  
Faculty of Health Science  
E46- 26 Old Main Building, Groote Schuur Hospital,  
Observatory, 7925  
Queries : Xolile Fula  
Tel : (021) 406-6492 Fax: 406-6411  
E-mail : Xfula@curie.uct.ac.za

14 March 2002

**REC REF: 086/2001**

Mr LW Viljoen  
Human Biology

Dear Mr Viljoen

THE EFFECT OF TENDON STIFFNESS ON STRETCH SHORTENING CYCLE  
PERFORMANCE

Thank you, for your letter to the Research Ethics Committee dated 13 March 2002.

*It is a pleasure to inform you that the addendum to the original above-mentioned study has been **approved** on the 13 March 2002*

Please quote the above Rec reference number in all correspondence

Signed by candidate

A/PROFESSOR CR SWANEPOEL  
CHAIRPERSON



**GENERAL INFORMATION FORM**  
**DEPARTMENT OF HUMAN BIOLOGY**  
*UNIVERSITY OF CAPE TOWN*

**GENERAL INFORMATION FORM WITH REGARDS TO  
EXERCISE TESTING AND BIOMEDICAL RESEARCH**

STRETCH-SHORTENING CYCLE AND MUSCULOTENDINOUS STIFFNESS TEST

**CLIENT NUMBER/CODE**

**COMPLETE NAME**

**POSTAL ADDRESS**

**Date of birth**

**Age**

**Telephone numbers**

(h)  
(w)  
(Cell)

**E-mail address**

**Sport**  
**(active participation in the last 12 months)**

**Gender**

Do you have any resistance training experience? ☐ Y ☐ N

If yes, please specify in the space provided for how many years and for what purpose the training was performed.


Do you have, or have you had, any musculoskeletal injuries? ☐ Y ☐ N

If yes to either of the above, please specify with details in the space provided.


For how many years have you been actively participating in this sport? Please specify at which level.


During the last 3 months, how many times on average, per week, have you trained?

1	2	3	4	5	6	7	8	9	10 or more
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Please list any illness, hospitalisation or surgical procedure within the past 2 years.




If there is any other information, which might be deemed important with regards to this study you would like to mention, please use the space below.


I, the undersigned, \_\_\_\_\_, state that all the information I have given in the above questionnaire is accurate and correct according my knowledge.

DATE:

--

NAME (SUBJECT):

--

SIGNATURE (SUBJECT):

--

DATE:

--

NAME (WITNESS):

--

SIGNATURE (WITNESS):

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DATE:

--

NAME (TESTER):

LW VILJOEN
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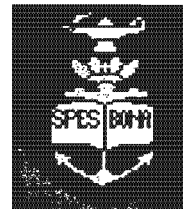
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*Once again thank you for your willingness to participate in this study and support the field of exercise science and biomedical research.*

Regards

Wayne Viljoen



## **INFORMED CONSENT FORM**

### **DEPARTMENT OF HUMAN BIOLOGY**

*UNIVERSITY OF CAPE TOWN*

## **GENERAL INFORMED CONSENT WITH REGARDS TO EXERCISE TESTING AND BIOLOGICAL RESEARCH**

### **1. EXPLANATION OF THE TESTS**

The MRC/UCT Research Unit for Exercise Science and Sports Medicine will be conducting a study focused on the test re-test repeatability of the Oscillation technique for measurement of muscle-tendon complex stiffness in both upper and lower body, using the newly developed NAMS Unit and its attachments (Zest Manufacturing PTY (Ltd), Cape Town, South Africa).

The testing will be over four days, with a 48-hour break in between days in order to ensure adequate recovery. The following tests will be performed:

#### **Day 1:**

- Weight and height
- Percentage body fat
- 1RM (one-repetition maximum) rebound bench press OR
- 1RM (one-repetition maximum) isometric reclined leg press
- Familiarisation session for the oscillation techniques for either upper OR lower body

#### **Day 2:**

- The Oscillation maximal stiffness test for the upper body OR
- The Oscillation maximal stiffness test for the lower body
- EMG (electromyography) will be used for all tests

#### **Day 3:**

- The Oscillation maximal stiffness test for the upper body OR
- The Oscillation maximal stiffness test for the lower body
- EMG (electromyography) will be used for all tests

#### **Day 4:**

- The Oscillation maximal stiffness test for the upper body OR
- The Oscillation maximal stiffness test for the lower body
- EMG (electromyography) will be used for all tests

## **2. ATTENDANT RISKS AND DISCOMFORTS**

It is of utmost importance that the subjects are injury free in the involved areas. If not, then this information should be shared with the tester beforehand and after consultation with a qualified medical doctor at the research unit, decided if the subject can participate in the study or not. Due to the nature of the testing, there is always the risk of muscle injury during the 1RM, MVC or oscillation tests if the subject does not warm-up adequately or is carrying an injury. This risk is however minimal if the subject follows the prescribed test protocol completely.

## **3. RESPONSIBILITIES OF THE PARTICIPANT**

It is important that the subjects follow the instructions of the tester completely throughout the testing time-period. For each testing day, the subjects should be completely rested and should not have trained at all for at least 24 hours before each testing day. This is of utmost importance for the repeatability measures of the NAMS Unit and Stiffness System.

## **4. BENEFITS TO BE EXPECTED**

Each subject will receive a brief summary of their test results regarding e.g. percentage body fat, maximal output of the upper body or maximal force output of the lower body, musculotendinous stiffness of the upper body or musculotendinous stiffness of the lower body. They will further have the knowledge that they have contributed to developing new scientific research in the field of exercise science and human biology.

## **5. INQUIRIES**

The subjects may feel free to ask any questions regarding the testing procedure and research at any time during, before or after the testing procedures.

## **6. FREEDOM OF CONSENT**

If subjects feel the need to withdraw from the study, they may feel free to do so at any time. Notice of this decision should however be given to the researcher involved.

I confirm that the above-mentioned tests have been thoroughly explained to me. I acknowledge that the personal information required by the researchers and those derived from the testing procedures will remain strictly confidential and no reference to my name will be revealed in any publication or statistical analysis. I have read this form and I understand the testing procedures that I will have to perform, my rights as a subject and the attendant risks, complications and discomforts. Knowing these risks, complications and discomforts, and having the opportunity to ask questions that have been answered to my satisfaction, I consent to participate in this study and offer my full cooperation.

**DATE:**

**NAME (SUBJECT):**

**SIGNATURE (SUBJECT):**

**DATE:**

**NAME (WITNESS):**

**SIGNATURE (WITNESS):**

**DATE:**

**NAME (TESTER):**

**SIGNATURE (TESTER):**

University of Cape Town



# **INFORMED CONSENT FORM**

## **DEPARTMENT OF HUMAN BIOLOGY**

### **UNIVERSITY OF CAPE TOWN**

## **GENERAL INFORMED CONSENT WITH REGARDS TO EXERCISE TESTING AND BIOLOGICAL RESEARCH**

### **1. EXPLANATION OF THE TESTS**

The MRC/UCT Research Unit of Exercise Science and Sports Medicine will be conducting a study focused on the test re-test repeatability of the stretch-shortening cycle testing techniques for measurement of stretch-shortening cycle or elastic ability in both upper and lower body, using the newly developed NAMS Unit and its attachments (Zest Manufacturing PTY (Ltd), Cape Town, South Africa).

The testing will be over four days, with a 48-hour break in between days in order to ensure adequate recovery. The following tests will be performed:

#### **Day 1:**

- Weight and height
- Percentage body fat
- Familiarisation session for the testing techniques

#### **Day 2:**

- MVC (maximum voluntary contraction) isometric bench press OR
- MVC (maximum voluntary contraction) isometric leg press
- The bench throw SSC test for the upper body OR
- The vertical jump SSC test for the lower body
- EMG (electromyography) will be used for all tests

#### **Day 3:**

- MVC (maximum voluntary contraction) isometric bench press OR
- MVC (maximum voluntary contraction) isometric leg press
- The bench throw SSC test for the upper body OR
- The vertical jump SSC test for the lower body
- EMG (electromyography) will be used for all tests

#### **Day 4:**

- MVC (maximum voluntary contraction) isometric bench press OR
- MVC (maximum voluntary contraction) isometric leg press
- The bench throw SSC test for the upper body OR
- The vertical jump SSC test for the lower body
- EMG (electromyography) will be used for all tests

## **2. ATTENDANT RISKS AND DISCOMFORTS**

It is of utmost importance that the subjects are injury free in the involved areas. If not, then this information should be shared with the tester beforehand and after consultation with a qualified medical doctor at the research unit, decided if the subject can participate in the study or not. Due to the nature of the testing, there is always the risk of muscle injury during the MVC and SSC tests if the subject does not warm-up adequately or is carrying an injury. This risk is however minimal if the subject follows the prescribed test protocol completely.

## **3. RESPONSIBILITIES OF THE PARTICIPANT**

It is important that the subjects follow the instructions of the tester completely throughout the testing time-period. For each testing day, the subjects should be completely rested and should not have trained at all for at least 24 hours before each testing day. This is of utmost importance for the repeatability measures of the NAMS Unit.

## **4. BENEFITS TO BE EXPECTED**

Each subject will receive a brief summary of their test results regarding body percentage body fat, maximal force output of the upper body or maximal force output of the lower body, elastic ability of the upper body or elastic ability of the lower body. They will further have the knowledge that they have contributed to developing new scientific research in the field of exercise science and human biology.

## **5. INQUIRIES**

The subjects may feel free to ask any questions regarding the testing procedure and research at any time during, before or after the testing procedures.

## **6. FREEDOM OF CONSENT**

If subjects feel the need to withdraw from the study, they may feel free to do so at any time. Notice of this decision should however be given to the researcher involved.

I confirm that the above-mentioned tests have been thoroughly explained to me. I acknowledge that the personal information required by the researchers and those derived from the testing procedures will remain strictly confidential and no reference to my name will be revealed in any publication or statistical analysis. I have read this form and I understand the testing procedures that I will have to perform, my rights as a subject and the attendant risks, complications and discomforts. Knowing these risks, complications and discomforts, and having the opportunity to ask questions that have been answered to my satisfaction, I consent to participate in this study and offer my full cooperation.

**DATE:**

**NAME (SUBJECT):**

**SIGNATURE (SUBJECT):**

**DATE:**

**NAME (WITNESS):**

**SIGNATURE (WITNESS):**

**DATE:**

**NAME (TESTER):**

**LW VILJOEN**

**SIGNATURE (TESTER):**

University of Cape Town



**INFORMED CONSENT FORM**  
**DEPARTMENT OF HUMAN BIOLOGY**  
*UNIVERSITY OF CAPE TOWN*

**GENERAL INFORMED CONSENT WITH REGARDS TO  
EXERCISE TESTING AND BIOLOGICAL RESEARCH**

**1. EXPLANATION OF THE TESTS**

The MRC/UCT Research Unit of Exercise Science and Sports Medicine will be conducting a study focused on the test modification of the Oscillation technique for measurement of muscle-tendon complex stiffness in the lower body, using the newly developed NAMS Unit and its attachments (Zest Manufacturing PTY (Ltd), Cape Town, South Africa).

The testing will be over two days, with a 48-hour break in between days in order to ensure adequate recovery. The following tests will be performed:

**Day 1:**

- Weight and height
- Percentage body fat
- Familiarisation session for the oscillation techniques for the lower body

**Day 2:**

- The modified Oscillation maximal stiffness test for the lower body
- EMG (electromyography) will be used

**2. ATTENDANT RISKS AND DISCOMFORTS**

It is of utmost importance that the subjects are injury free in the involved areas. If not, then this information should be shared with the tester beforehand and after consultation with a qualified medical doctor at the research unit, decided if the subject can participate in the study or not. Due to the nature of the testing, there is always the risk of muscle injury during the stiffness tests if the subject does not warm-up adequately or is carrying an injury. This risk is however minimal if the subject follows the prescribed test protocol completely.



### **3. RESPONSIBILITIES OF THE PARTICIPANT**

It is important that the subjects follow the instructions of the tester completely throughout the testing time-period. For each testing day, the subjects should be completely rested and should not have trained at all for at least 24 hours before each testing day. This is of utmost importance for the repeatability measures of the NAMS Unit and Stiffness System.

### **4. BENEFITS TO BE EXPECTED**

Each subject will receive a brief summary of their test results regarding e.g. percentage body fat, and musculotendinous stiffness of the lower body. They will further have the knowledge that they have contributed to developing new scientific research in the field of exercise science and human biology.

### **5. INQUIRIES**

The subjects may feel free to ask any questions regarding the testing procedure and research at any time during, before or after the testing procedures.

### **6. FREEDOM OF CONSENT**

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**DATE:**

**NAME (SUBJECT):**

**SIGNATURE (SUBJECT):**

**DATE:**

**NAME (WITNESS):**

**SIGNATURE (WITNESS):**

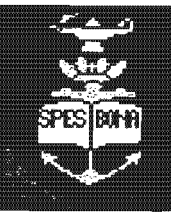
**DATE:**

**NAME (TESTER):**

**LW VILJOEN**

**SIGNATURE (TESTER):**

University of Cape Town



**INFORMED CONSENT FORM**  
**DEPARTMENT OF HUMAN BIOLOGY**  
*UNIVERSITY OF CAPE TOWN*

**GENERAL INFORMED CONSENT WITH REGARDS TO  
EXERCISE TESTING AND BIOLOGICAL RESEARCH**

**1. EXPLANATION OF THE TESTS**

The MRC/UCT Research Unit of Exercise Science and Sports Medicine will be conducting a study focused on determining the relationship between musculotendinous stiffness and stretch-shortening cycle or elastic ability in either upper or lower body, using the newly developed NAMS Unit and its attachments (Zest Manufacturing PTY (Ltd), Cape Town, South Africa).

The testing will be over three days, with a 48-hour break in between days in order to ensure adequate recovery. The following tests will be performed:

**Day 1:**

- Weight and height
- Percentage body fat
- MVC (maximum voluntary contraction) isometric leg press OR
- MVC (maximum voluntary contraction) isometric bench press
- Familiarisation session for the oscillation and SSC techniques

**Day 2:**

- The Oscillation maximal stiffness test for the lower body OR
- The Oscillation maximal stiffness test for the upper body
- EMG (electromyography) will be used for all tests

**Day 3:**

- MVC (maximum voluntary contraction) isometric leg press OR
- MVC (maximum voluntary contraction) isometric bench press
- The vertical jump SSC test for the lower body OR
- The bench throw SSC test for the upper body
- EMG (electromyography) will be used for all tests

## **2. ATTENDANT RISKS AND DISCOMFORTS**

It is of utmost importance that the subjects are injury free in the involved areas. If not, then this information should be shared with the tester beforehand and after consultation with a qualified medical doctor at the research unit, decided if the subject can participate in the study or not. Due to the nature of the testing, there is always the risk of muscle injury during the MVC, SSC and oscillation tests if the subject does not warm-up adequately or is carrying an injury. This risk is however minimal if the subject follows the prescribed test protocol completely.

## **3. RESPONSIBILITIES OF THE PARTICIPANT**

It is important that the subjects follow the instructions of the tester completely throughout the testing time-period. For each testing day, the subjects should be completely rested and should not have trained at all for at least 24 hours before each testing day. This is of utmost importance for the accuracy of the measures on the NAMS Unit.

## **4. BENEFITS TO BE EXPECTED**

Each subject will receive a brief summary of their test results regarding percentage body fat, maximal force output of the upper body or maximal force output of the lower body, musculotendinous stiffness of the upper body or musculotendinous stiffness of the lower body, elastic ability of the upper body or elastic ability of the lower body. They will further have the knowledge that they have contributed to developing new scientific research in the field of exercise science and human biology.

## **5. INQUIRIES**

The subjects may feel free to ask any questions regarding the testing procedure and research at any time during, before or after the testing procedures.

## **6. FREEDOM OF CONSENT**

If subjects feel the need to withdraw from the study, they may feel free to do so at any time. Notice of this decision should however be given to the researcher involved.

I confirm that the above-mentioned tests have been thoroughly explained to me. I acknowledge that the personal information required by the researchers and those derived from the testing procedures will remain strictly confidential and no reference to my name will be revealed in any publication or statistical analysis. I have read this form and I understand the testing procedures that I will have to perform, my rights as a subject and the attendant risks, complications and discomforts. Knowing these risks, complications and discomforts, and having the opportunity to ask questions that have been answered to my satisfaction, I consent to participate in this study and offer my full cooperation.

**DATE:**

**NAME (SUBJECT):**

**SIGNATURE (SUBJECT):**

**DATE:**

**NAME (WITNESS):**

**SIGNATURE (WITNESS):**

**DATE:**

**NAME (TESTER):**

**SIGNATURE (TESTER):**

University of Cape Town